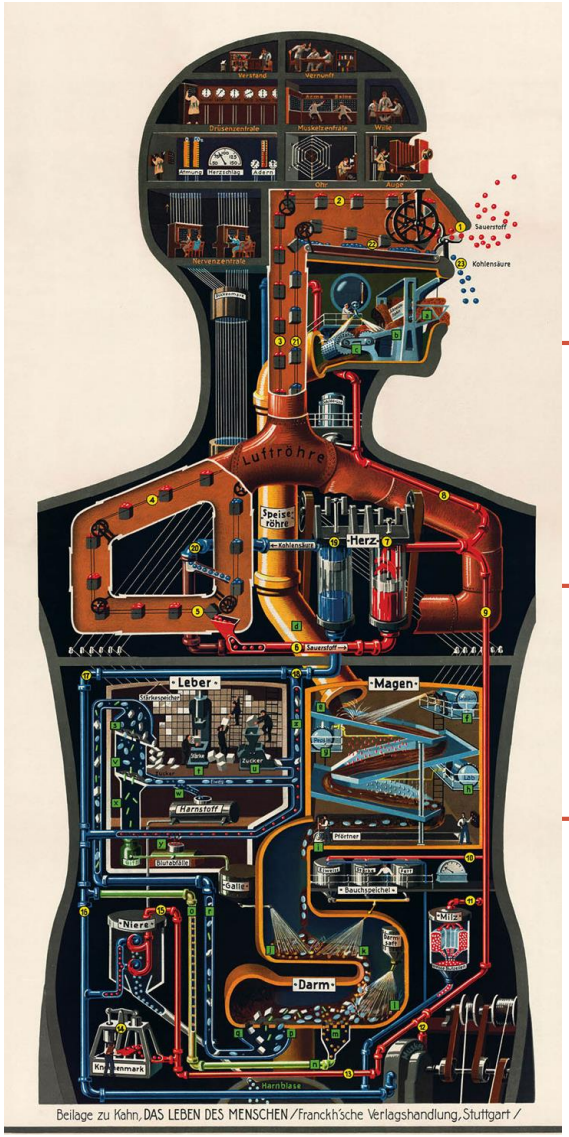


Fritz Kahn (1888 – 1968)

BIOMECHANICS

4: Numerical and Experimental Methods in Biomechanics

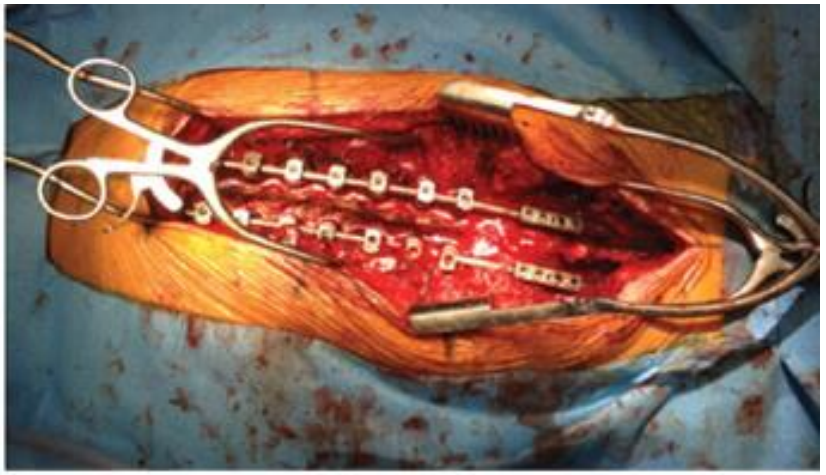
7ο εξάμηνο

Σχολή Μηχανολόγων Μηχανικών ΕΜΠ

Διδάσκων:
Michael Neidlin

Why to model?

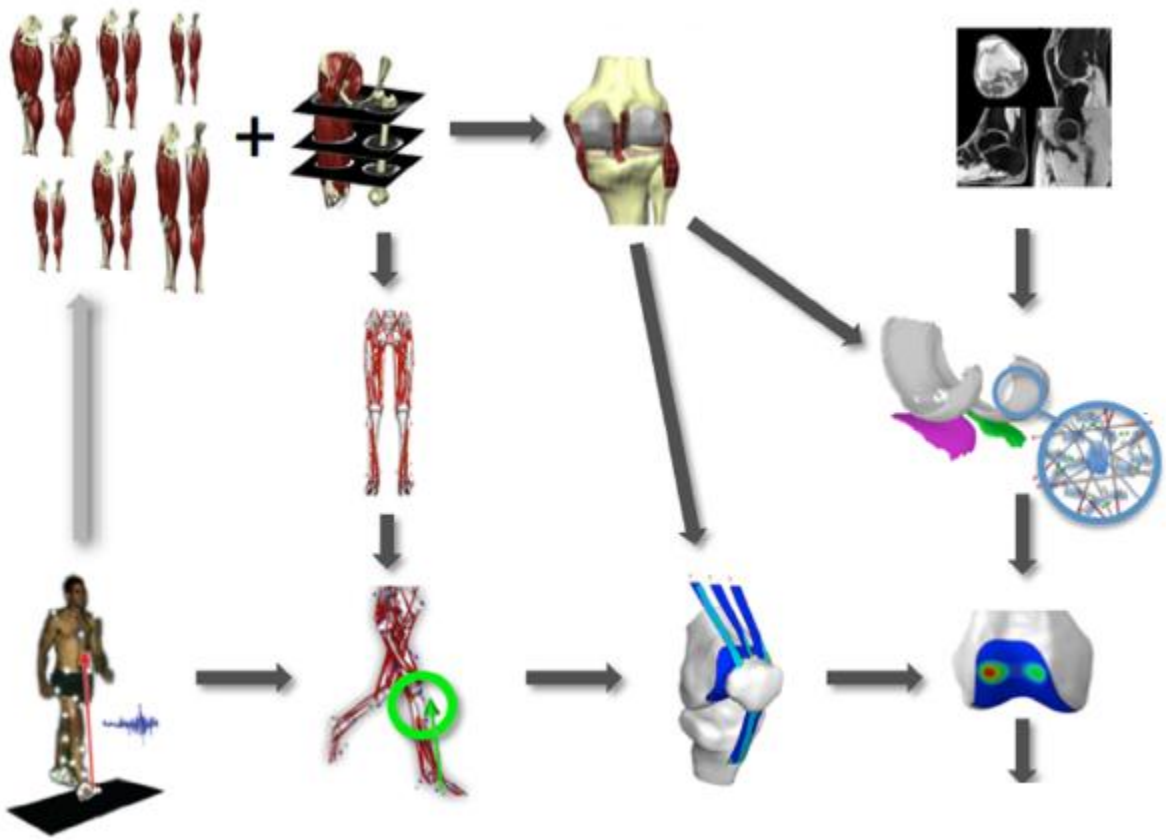
Minimally Invasive Spine Surgery



Traditional Open Spine Surgery

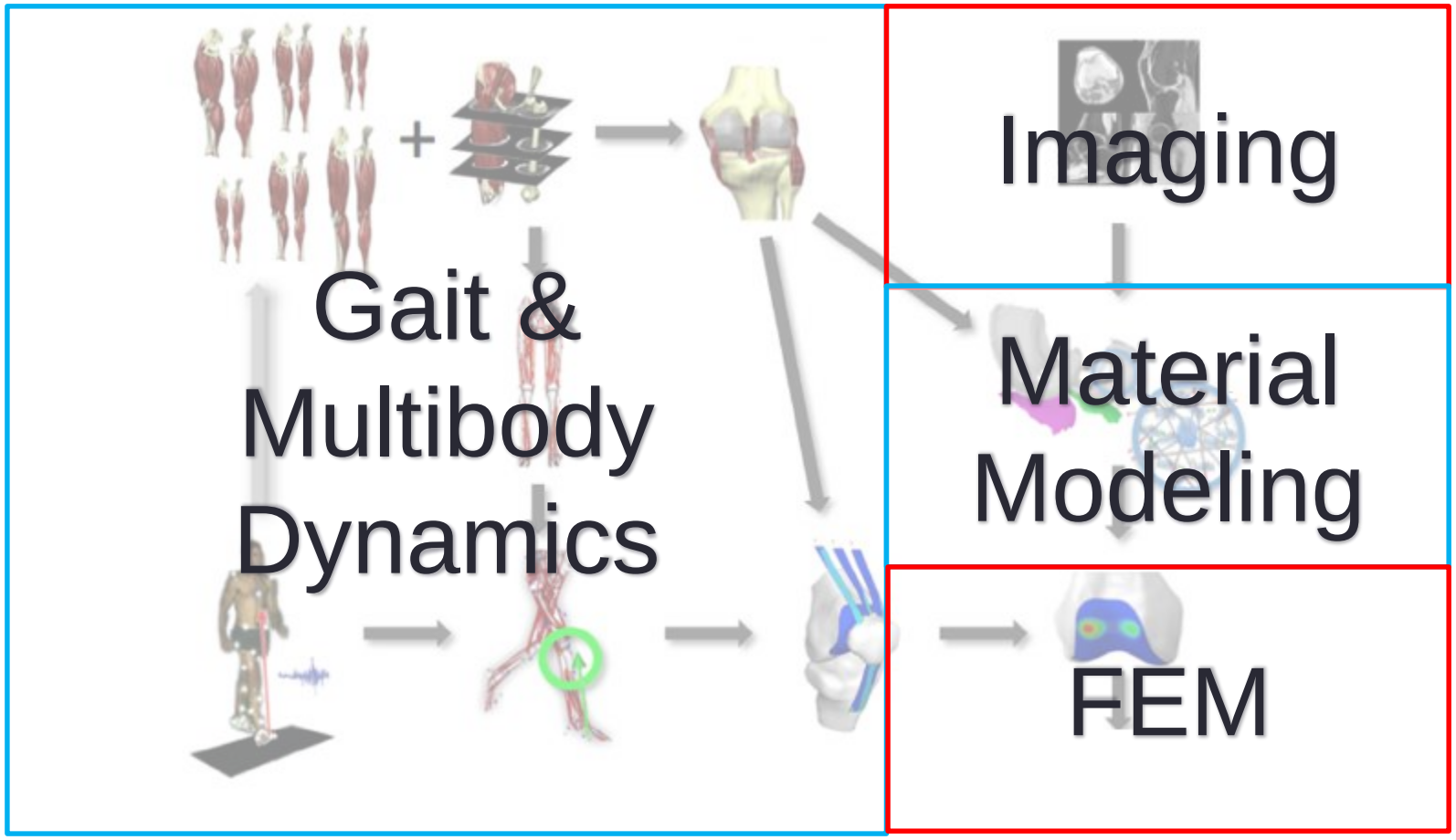
Improve conservative and surgical treatment of joint diseases

How to model?



Multiscale consideration of human movement. From human level to tissue level. (Cellular and molecular level also exists)

What to model?



Content

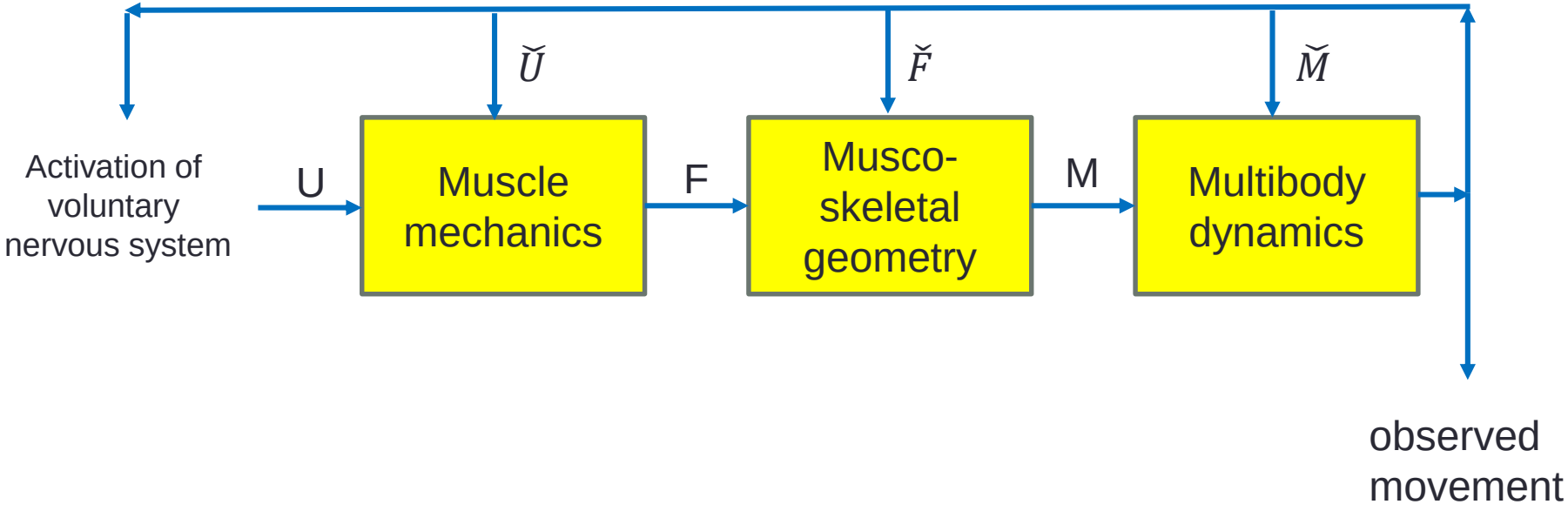
- Muscular and skeletal movement ➔ Joint forces/dynamics
- Material testing and modeling ➔ Material properties
- **FEM simulations** and validation approaches ➔ Stress/strain
- **Imaging and segmentation** ➔ Geometry extraction

- Information for the practical course (part I)

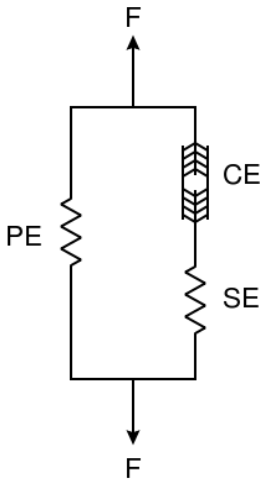
Important remark: It is not possible to present all the methods in biomechanics.

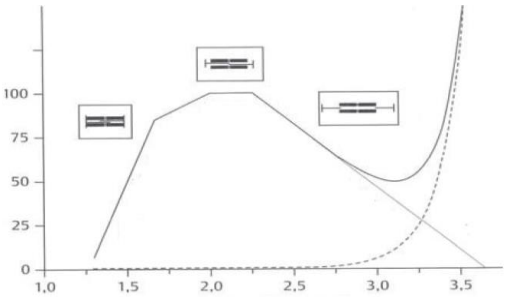
Everything on tissue/cell level will be presented in L5 and L6 (TE and cell mechanics)

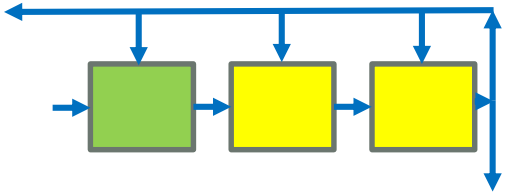
Control loop of human movement



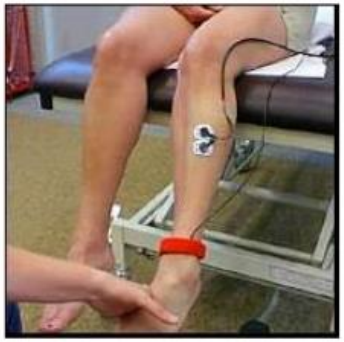
Muscle mechanics (Hill-Model)

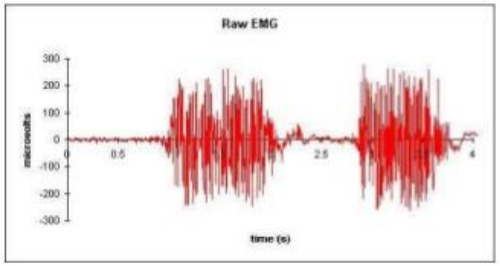






Electro**m**yo**g**raphy

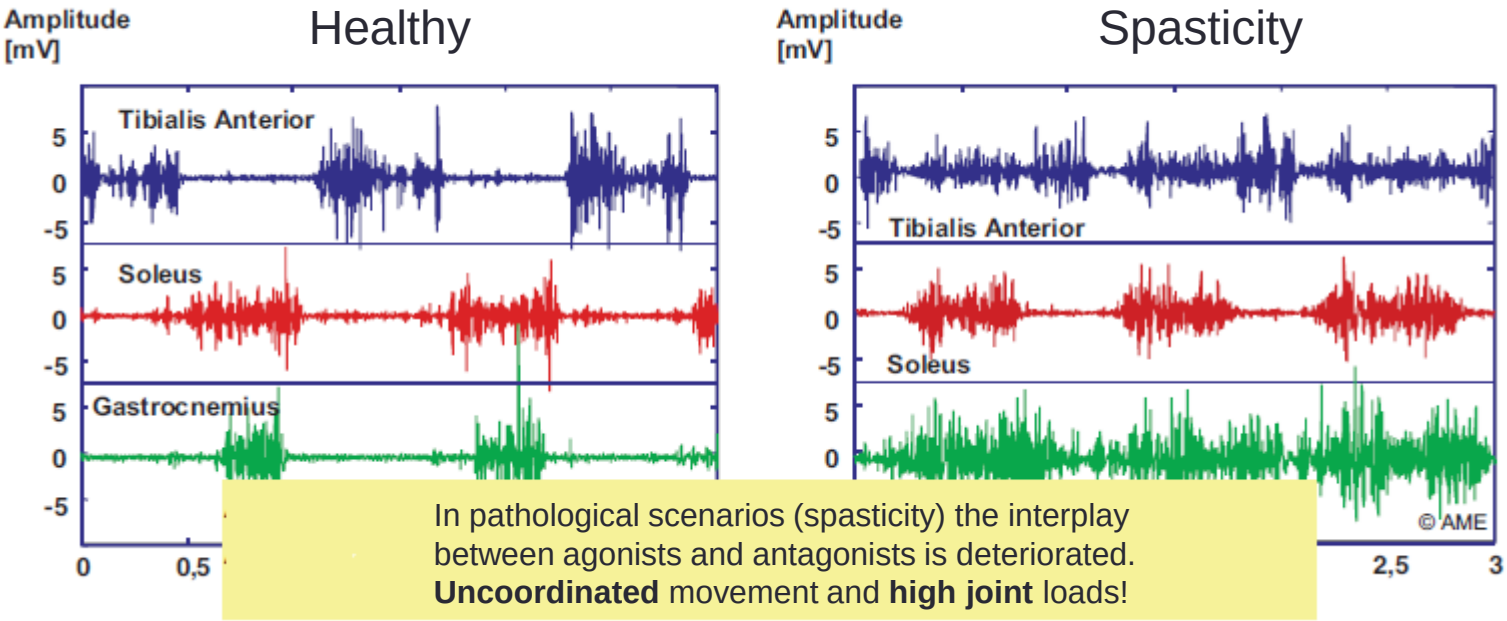




Signal processing (Filtering, FFT, Thresholding..)

Which muscle is activated?
At what time point?
What is the force **F**?

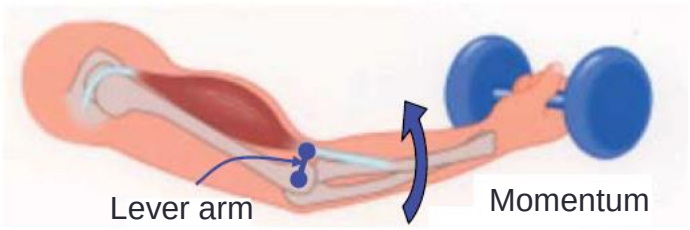
EMG example

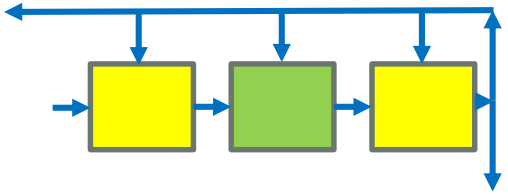


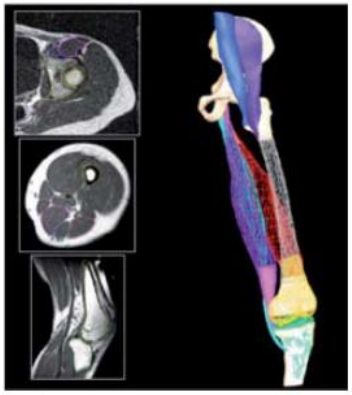
In pathological scenarios (spasticity) the interplay between agonists and antagonists is deteriorated. **Uncoordinated** movement and **high joint** loads!

EMG + gait monitoring can be used for diagnosis and therapy

Musco-skeletal geometry



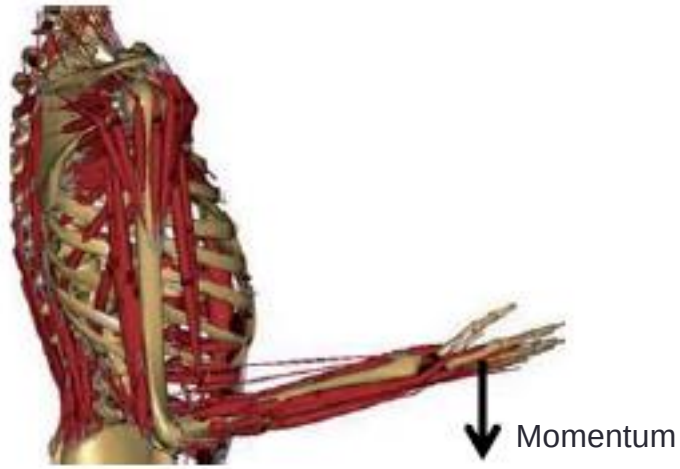




Starting points and end points of muscles
Bone size and orientation

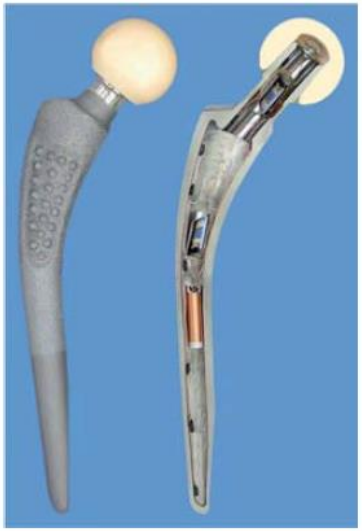
Imaging (difficult), anatomical knowledge, generic models

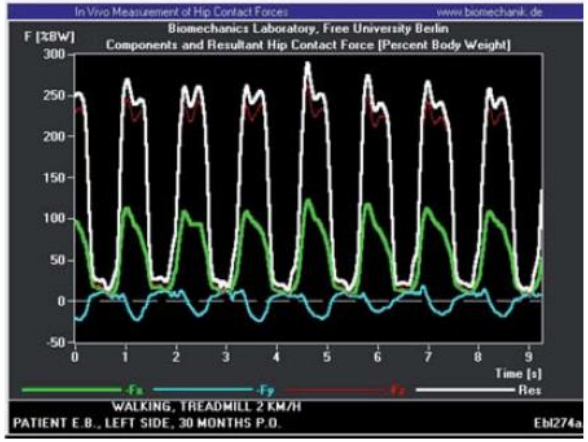
Only when studying individual joints

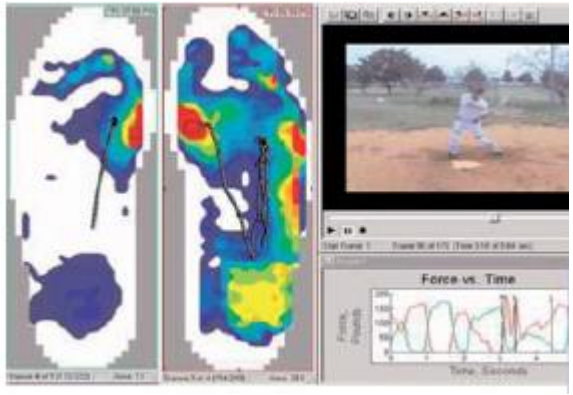


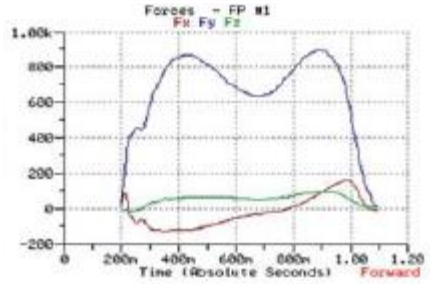
Measurement of forces

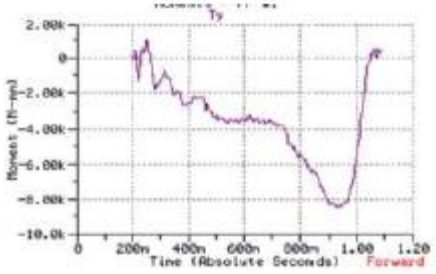
In vivo force measurement is invasive ➔ Instrumented implants or pressure indicating films



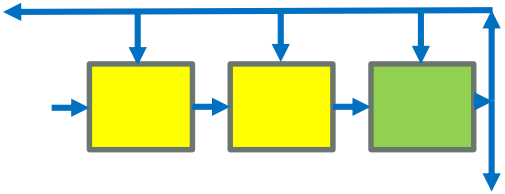






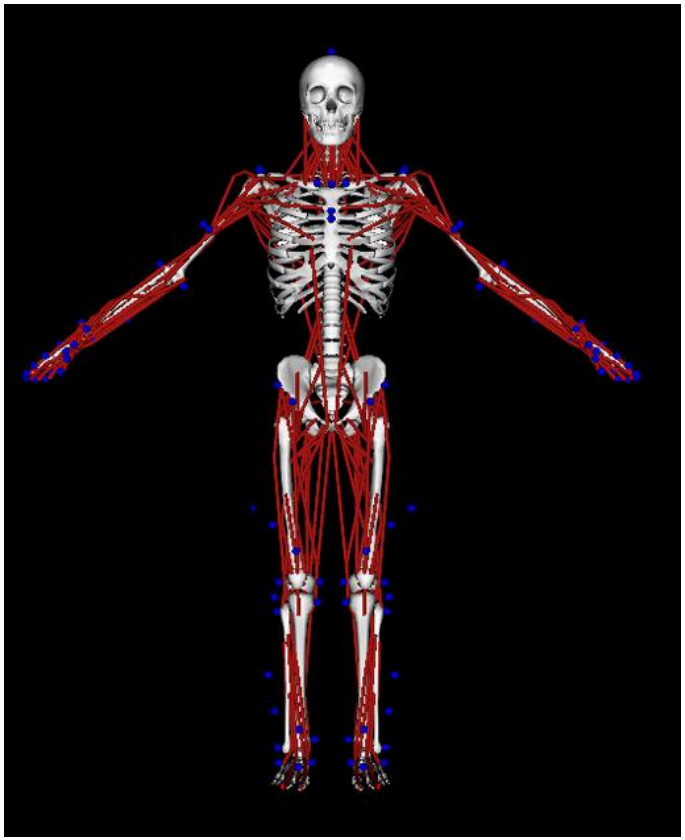


Multi-body dynamics



OpenSim

Movement
and moments
at joints



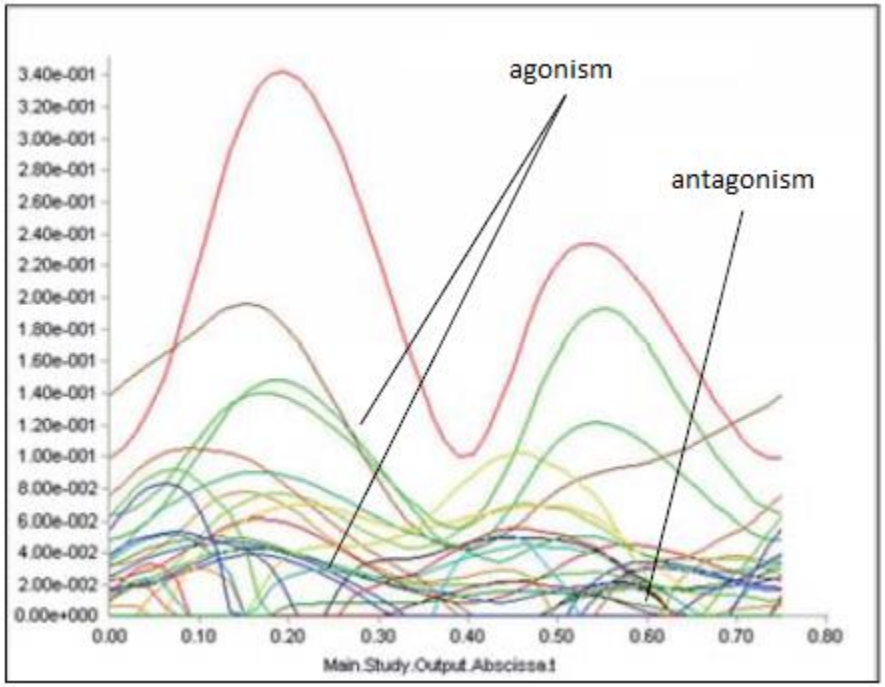
Activation profiles

Movement
of the entire body

Muscle forces

Inverse problem ➔ fitting to EMG data and movement adaptation until $EMG_{exp}=EMG_{sim}$

Complex model ➔ under-constrained optimization problem



Complex models (which **are not** necessarily better than simpler models) require further constraints to allow a **unique** solution.

Energy minimization of muscle activity

Joint loading minimization

Movement duration minimization

VALIDATION

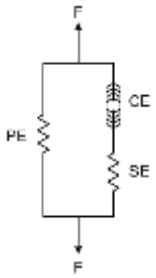
3D movement tracking

Validation of simulations

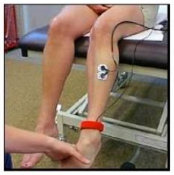
Providing boundary conditions
for simulations

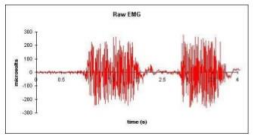
In vivo studies (pathologies,
rehabilitation, device design)
Connection
to other measurements (EMG)

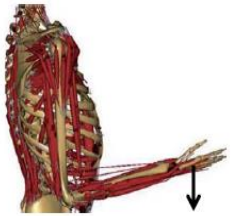
Summary - Gait



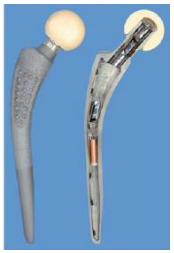
Muscle mechanics

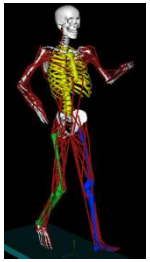






Musco-skeletal geometry





Multibody dynamics

Only the combination of experiment and simulation together with cross-validation can deliver a reliable and useful model!

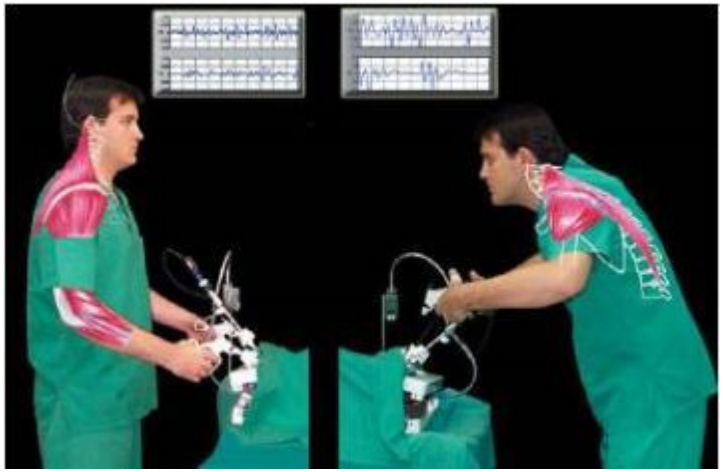
Examples : Pathologies - Stroke

https://www.youtube.com/watch?v=ecWBdL6-y6Q

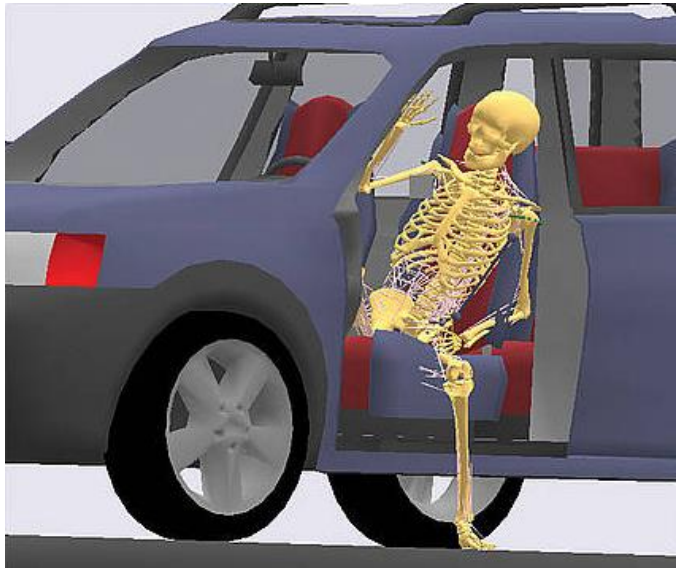
Examples : Ergonomics

Workplace

Surgery



Everyday Life

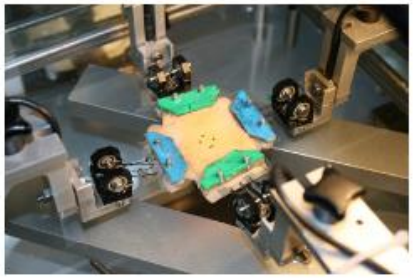


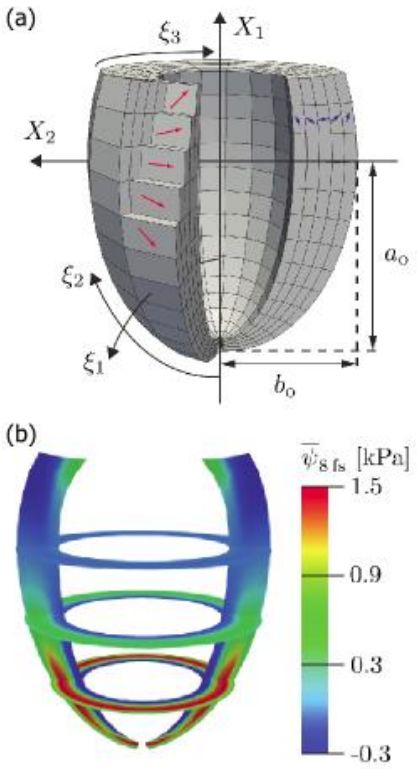
Examples: Sport biomechanics

https://www.youtube.com/watch?v=7u2Rqm5fXus

Questions?

Material testing and modeling – Why?





Get insight into the mechanical properties of tissues (devices, surgeries, general understanding...)

in vitro testing

Answer the scientific question

Quantify material parameters with **fitting of constitutive material models**

Implement in FE environment to get further insight (cartilage stress in full joint geometry)

Properties of biological tissues

L3: Biological tissues are

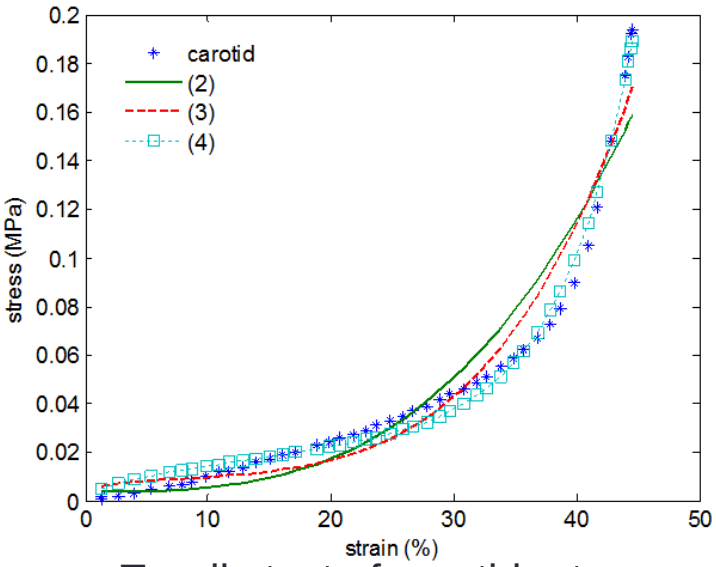
- non-linear elastic (**hyperelastic**)
- stiffness depends on loading direction (**anisotropic**)
- stiffness depends on loading velocity (**viscoelastic**)
- creep/relaxation/hysteresis (**viscoelastic**)

In special cases, e.g. cartilage

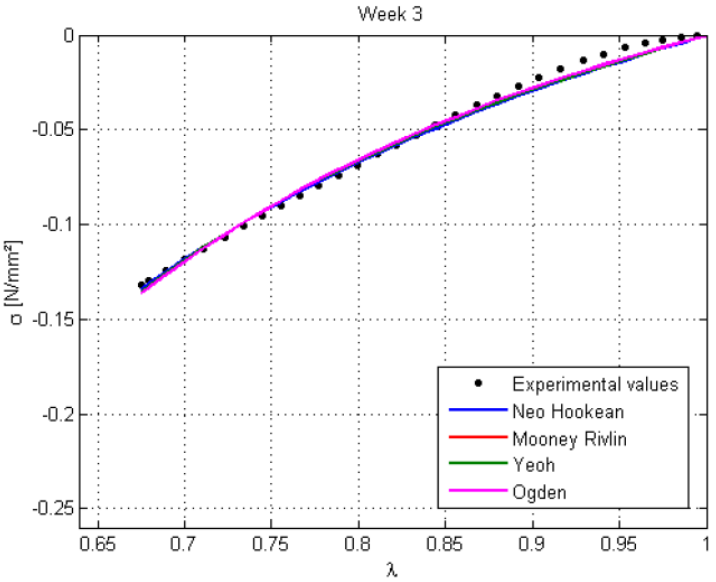
- fluid embedded in solid matrix (bi or tri-phasic)

- Long-term observation includes **remodeling**

Hyperelasticity



Tensile test of carotid artery



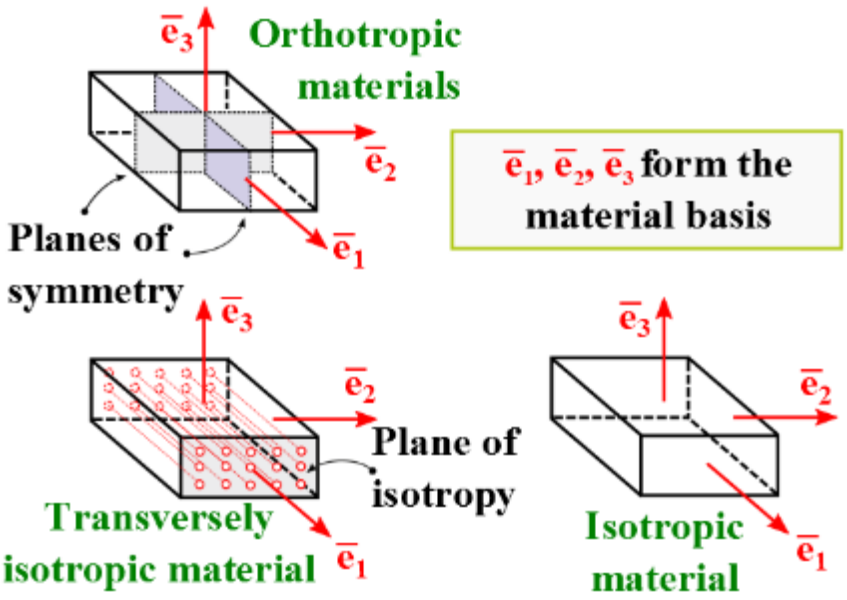
Compression test of cartilage

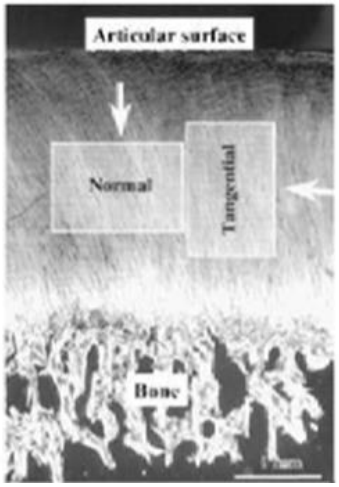
Non-linear stress-strain behavior of an **elastic** material (no plastic deformation)

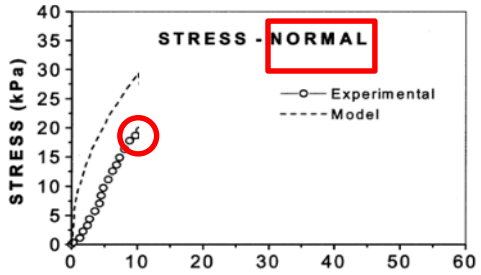
Described by a **strain energy density** function

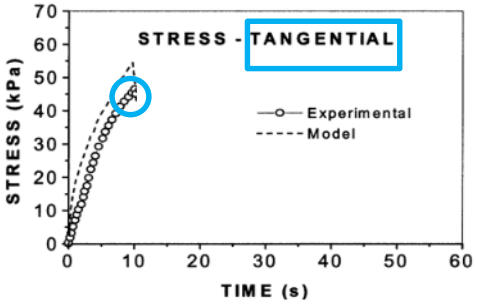
First developed for elastomers and polymers

Anisotropy









Compression tests of bone

Stress-strain behavior depends on loading direction (**anisotropy**)

Orthotropy (3 different Young's moduli) and **transversal isotropy** (2 different Young's moduli) are the most common forms of **directed anisotropy**

Loading and remodeling defines material directions (**Wolff's Law**!)

Identification of material parameters from in vitro data

Decided what we want to know : anisotropic properties (E_1 and E_2) of tissue

Did our experiments : tension/compression tests is axial direction 1 and 2

Choose a mathematical representation of our material model

Fit our material model to the data to derive the Young's modulus

Continuum mechanics of soft tissues

Remark: Normally some “new” mathematics needed. Tensor algebra and analysis (E.g. Book by Prof. Itskov from RWTH Aachen University) for the extremely interested student. As well as books by G.A. Holzapfel and G.W. Ogden.
In application the following is done by software (Abaqus FEM). What follows is a **very** short presentation of the steps from material model choice to parameter identification.

Definitions – Material deformation

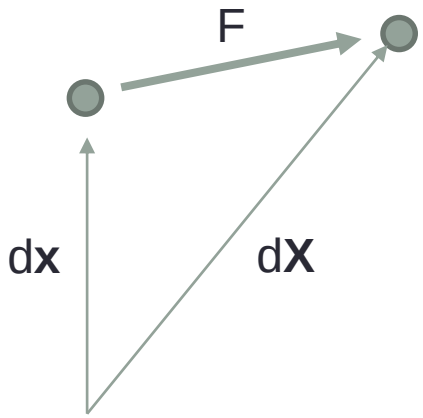
$$d\mathbf{x} = \mathbf{F} d\mathbf{X}$$

d**x** is the deformed configuration

d**X** is the reference configuration

F maps d**X** on d**x**.

$$[F_j^i] = [\frac{\partial x^i}{\partial X^j}] = \begin{bmatrix} \frac{\partial x^1}{\partial X^1} & \frac{\partial x^1}{\partial X^2} & \frac{\partial x^1}{\partial X^3} \\ \frac{\partial x^2}{\partial X^1} & \frac{\partial x^2}{\partial X^2} & \frac{\partial x^2}{\partial X^3} \\ \frac{\partial x^3}{\partial X^1} & \frac{\partial x^3}{\partial X^2} & \frac{\partial x^3}{\partial X^3} \end{bmatrix}$$



$$\lambda = \sqrt{\frac{\|d\mathbf{x}\|^2}{\|d\mathbf{X}\|^2}}$$

Stretch, e.g. λ=1.1 if elongation from 1m to 1.1m

$$\mathbf{C} = \mathbf{F}^T\mathbf{F}$$

Right **Cauchy-Green Tensor**

$$I_1 = \operatorname{tr}\mathbf{C},$$

$$I_2 = \frac{1}{2}[(\operatorname{tr}\mathbf{C})^2 - \operatorname{tr}\mathbf{C}^2],$$

$$I_3 = \det\mathbf{C},$$

Invariants do not change during deformation

Material deformation summary

Cauchy-Green tensor **C** describes material deformation

For every state of the material the Cauchy-Green tensor can be computed.

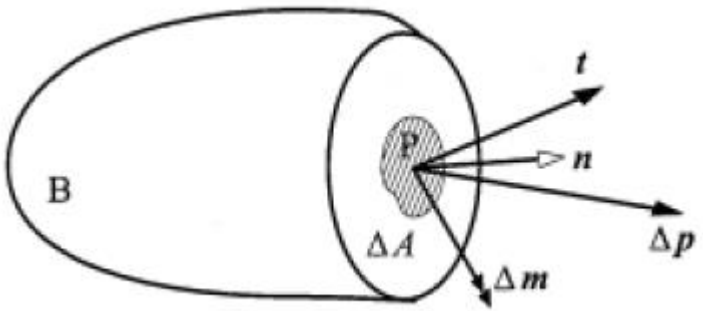
Invariants are calculated from **C.** Will be important in the next steps

Material strain (ε) is expressed with **Green-Lagrange** Strain Tensor **E**.

$$\mathbf{E} = \frac{1}{2}(\mathbf{C} - \mathbf{I})$$

Identity tensor

Definitions – Potatoes & Material stresses



Body **B** experiences after separation a force **Δp** on an area **ΔA**

Continuum mechanics postulate:
The limit exists and is final **t=dp/dA**!

We define a Cauchy stress vector **t**:

$$\boldsymbol{t} = \lim_{A \to 0} \frac{\Delta p}{\Delta A}$$

In the last step we map the unit vector **n** onto the stress vector **t** by the Cauchy stress tensor **σ!**

$$\boldsymbol{t} = \boldsymbol{\sigma} \boldsymbol{n} \qquad \boldsymbol{\sigma} = \begin{bmatrix} \sigma_{11} & \sigma_{12} & \sigma_{13} \\ \sigma_{21} & \sigma_{22} & \sigma_{23} \\ \sigma_{31} & \sigma_{32} & \sigma_{33} \end{bmatrix}$$

Normal stresses
Shear stresses

Definitions II – Stress Tensors

$$d\mathbf{p} = \boldsymbol{t} dA$$

Infinitesimal force on infinitesimal **deformed** area

$$\mathbf{P} = J\boldsymbol{\sigma}\mathbf{F}^{-T}$$

First Piola Kirchhoff stress tensor relates the stress to the **initial** configuration

J= det **C** = relative volume change J=1 if incompressible material

$$\mathbf{S} = J\mathbf{F}^{-1}\boldsymbol{\sigma}\mathbf{F}^{-T}$$

Second Piola Kirchhoff stress tensor relates the stress to the actual configuration

$$\boldsymbol{\sigma} = \mathbb{C} : \boldsymbol{\epsilon}$$

Generalized Hooke's Law

$$\mathbb{C} = 2\frac{\partial \mathbf{S}}{\partial \mathbf{C}} = \frac{\partial \mathbf{S}}{\partial \mathbf{E}}$$

Derivation of Stress Tensor **S** w.r.t. Strain Tensor **E**

Compliance Tensor C (4. order tensor) includes material properties (E,G,v)

Summary Continuum Mechanics Crash Course

All deformation can be expressed by Cauchy-Green tensor **C**

All stresses can be expressed by Piola-Kirchhoff tensor **P** or **S**, respectively

Stress and strain are connected with **Hooke's Law** via

$$\boldsymbol{\sigma} = \mathbb{C} : \boldsymbol{\epsilon}$$

$$\boldsymbol{\epsilon} = \mathbb{S} : \boldsymbol{\sigma}$$

$$\begin{bmatrix} \epsilon_{11} \\ \epsilon_{22} \\ \epsilon_{33} \\ \epsilon_{23} \\ \epsilon_{13} \\ \epsilon_{12} \end{bmatrix} = \begin{bmatrix} \frac{1}{E} & \frac{-\nu}{E} & \frac{-\nu}{E} & 0 & 0 & 0 \\ \frac{-\nu}{E} & \frac{1}{E} & \frac{-\nu}{E} & 0 & 0 & 0 \\ \frac{-\nu}{E} & \frac{-\nu}{E} & \frac{1}{E} & 0 & 0 & 0 \\ 0 & 0 & 0 & \frac{1}{2G} & 0 & 0 \\ 0 & 0 & 0 & 0 & \frac{1}{2G} & 0 \\ 0 & 0 & 0 & 0 & 0 & \frac{1}{2G} \end{bmatrix} \begin{bmatrix} \sigma_{11} \\ \sigma_{22} \\ \sigma_{33} \\ \sigma_{23} \\ \sigma_{13} \\ \sigma_{12} \end{bmatrix}$$

Inverse of compliance tensor for an isotropic material

Hyperelastic material defined by **strain energy function**

$$\Psi = \bar{\Psi}(I_1, I_2, I_3)$$

Strain energy function defined by invariants (deformation)
This is the **constitutive model provided** for fitting.

$$\mathbf{S} = 2\frac{\partial \Psi}{\partial \mathbf{C}}$$

Stress calculated from strain energy function ➔ connection between material model, stress and strain!

Neo-Hookean model (isotropic)

$$\Psi = C_1(I_1 - 3)$$

Material modeling procedure:

1. Do material tests
2. Calculate **C** from the deformation
3. Define $\mathbf{P}_{exp}$ or $\mathbf{S}_{exp}$ from force measurements
4. Derive $\mathbf{P}_{mod}$ or $\mathbf{S}_{mod}$ from **ψ**
5. Parameter fitting $\mathbf{P}_{exp}/\mathbf{P}_{mod}$ or $\mathbf{S}_{exp}/\mathbf{S}_{mod}$
6. Calculate E/G/v from fitted model

Material modeling example – uniaxial compression

$$\mathbf{F} = \begin{bmatrix} \lambda_1 & 0 & 0 \\ 0 & \lambda_2 & 0 \\ 0 & 0 & \lambda_3 \end{bmatrix}$$

incompressibility $\lambda_2 = \lambda_3 = \frac{1}{\sqrt{\lambda}}$
+
unconfined deformation

$$\mathbf{F} = \begin{bmatrix} \lambda & 0 & 0 \\ 0 & \frac{1}{\sqrt{\lambda}} & 0 \\ 0 & 0 & \frac{1}{\sqrt{\lambda}} \end{bmatrix}$$

$$\mathbf{C} = \begin{bmatrix} \lambda^2 & 0 & 0 \\ 0 & \frac{1}{\lambda} & 0 \\ 0 & 0 & \frac{1}{\lambda} \end{bmatrix}$$

$$I_1 = \lambda^2 + \frac{2}{\lambda},$$
$$I_2 = 2\lambda + \frac{1}{\lambda^2},$$
$$I_3 = 1.$$

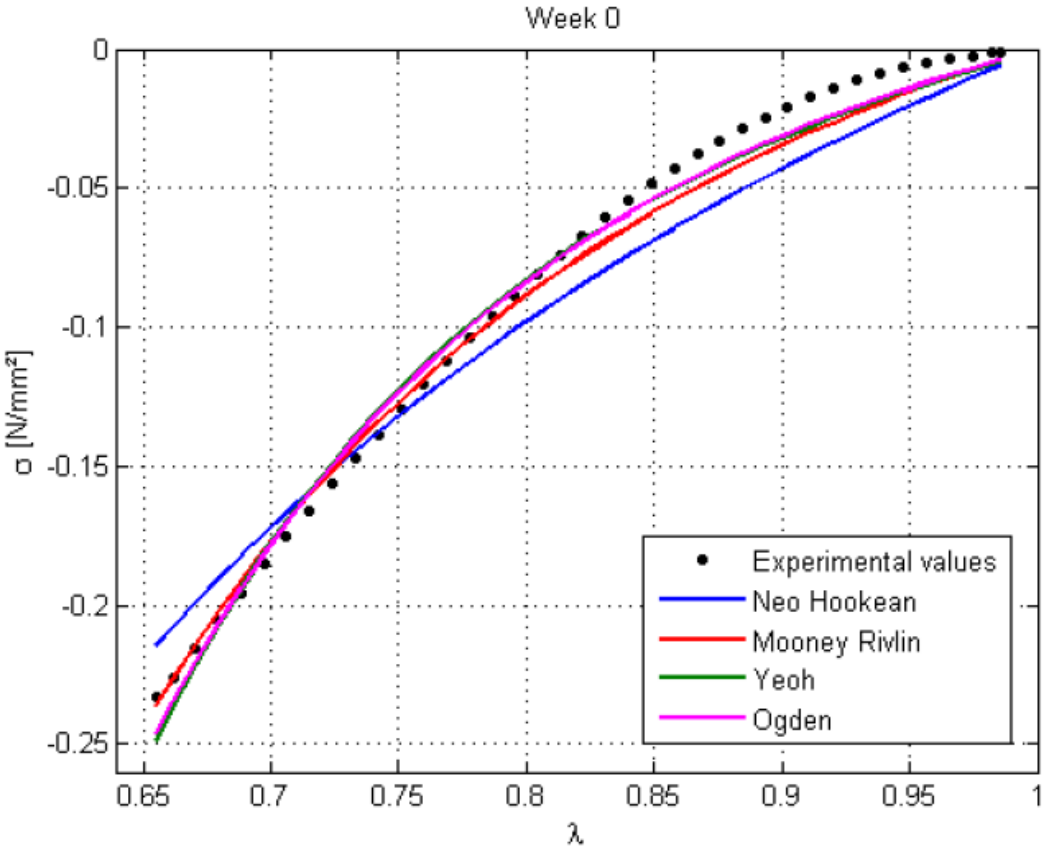
Invariants depend on the deformation conditions

$$\Psi = C_1(I_1 - 3)$$

$$P_{11} = \frac{d\Psi}{d\lambda} = \frac{d\Psi}{dI_1}\frac{dI_1}{d\lambda} = 2C_1(\lambda - \frac{1}{\lambda^2})$$

Stress in the first direction (output of compression tester)

Material modeling example – uniaxial compression



	Neo Hookean		Mooney Rivlin		Yeoh		Ogden	
	RMSE	*E*[MPa]	*RMSE*	*E*[MPa]	*RMSE*	*E*[MPa]	*RMSE*	*E*[MPa]
Week 0	0.0163	0.38	0.0087	0.276	0.0082	0.276	0.0078	0.246

Strain energy functions - This is what you select in FEM

Mooney Rivlin Model

$$\Psi = C_1(I_1 - 3) + C_2(I_2 - 3)$$

Ogden Model

$$\Psi = \sum_{r=1}^{s} \frac{\mu_r}{\alpha_r}(\lambda_1^{\alpha_r} + \lambda_2^{\alpha_r} + \lambda_3^{\alpha_r} - 3)$$

Holzapfel Model

$$\psi = \mu(I_1 - 3) + \frac{k_1}{k_2}(\exp\{k_2[(1-\rho)(I_1-3)^2 + \rho(I_4-1)^2]\} - 1)$$

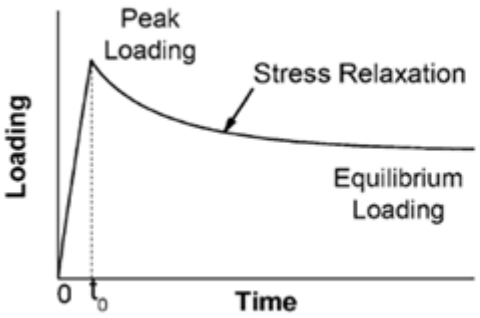
Isotropic matrix

Fiber-reinforcement,
I_4 depends on fiber direction ➔ anisotropy

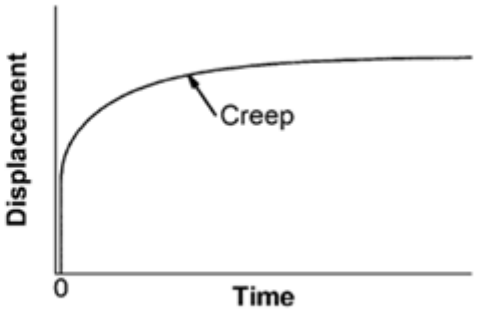
Very popular model for biological tissues. Included in common FE solvers like ABAQUS, ANSYS, Comsol...

Viscoelasticity

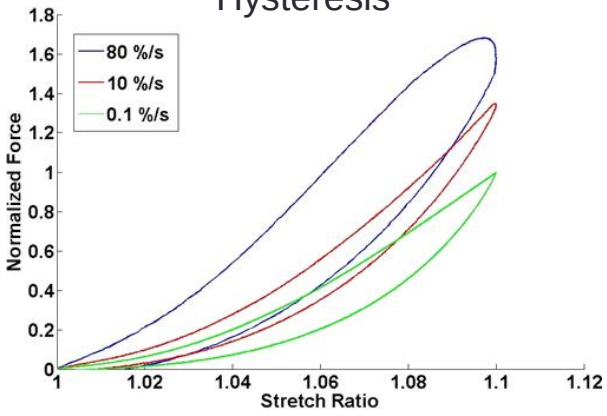
Relaxation

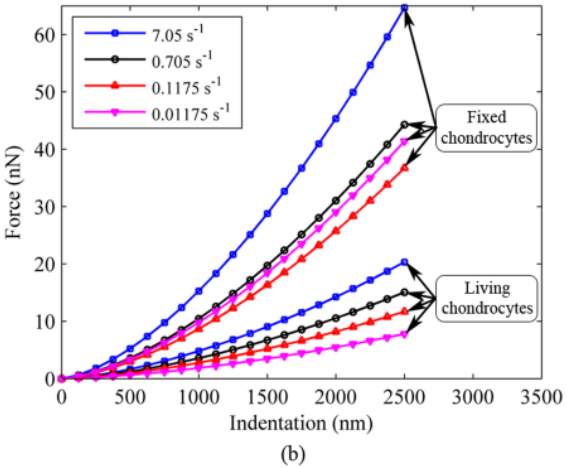


Creep



Hysteresis





Velocity-dependence

Energy dissipation during material deformation

Time-dependent stress-strain response

Included as an additional term in **material models**

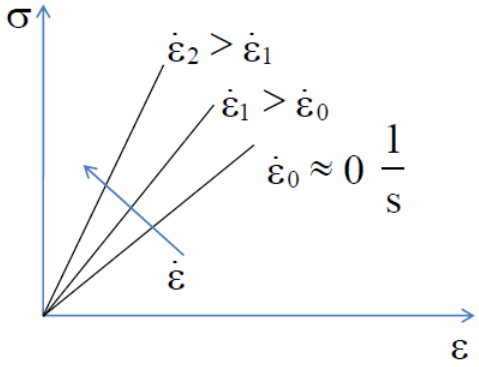
Viscosity – separate consideration

$$\sigma = E \cdot \varepsilon + a \cdot \dot{\varepsilon}^{b}$$

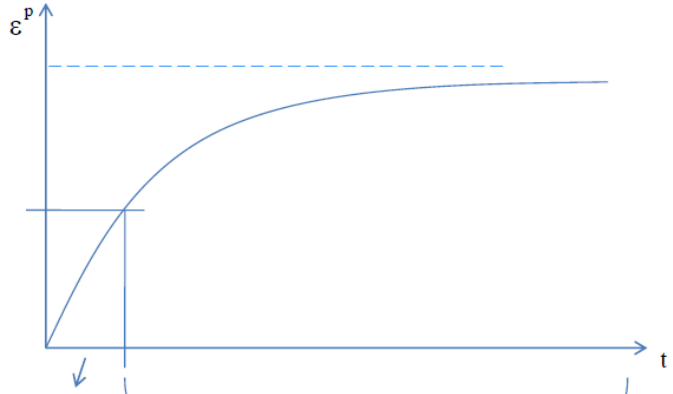
elastic part

viscous part

a,b from experiments



Creep



initial deformation

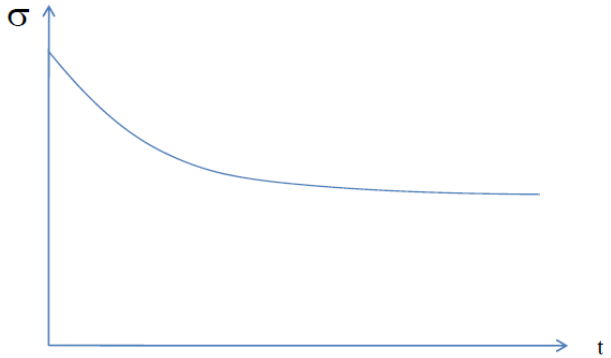
deformation increases with time

$$\dot{\varepsilon}^p = \left(\frac{\sigma}{K}\right)^n$$

$$\varepsilon^p = \left(\frac{\sigma}{K}\right)^n * t + \varepsilon_0^p$$

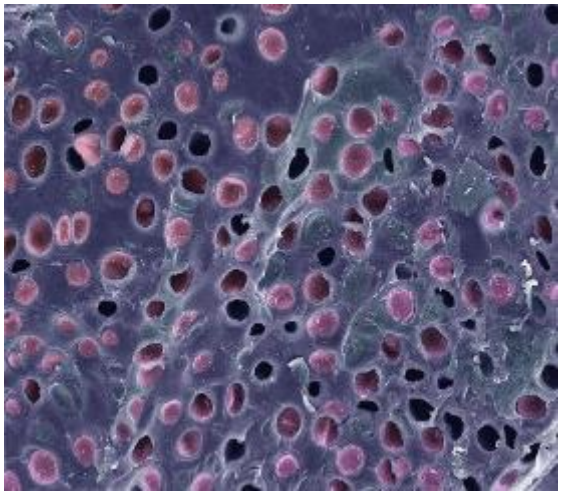
linear

Stress Relaxation

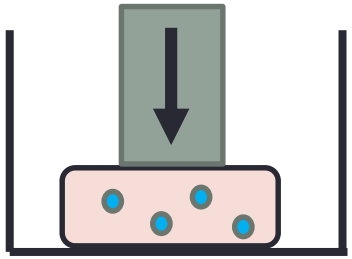


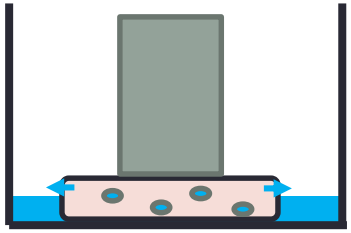
$$\sigma = A * e^{-Bt}$$

Poroviscoelasticity of cartilage



More in Cell mechanics lecture



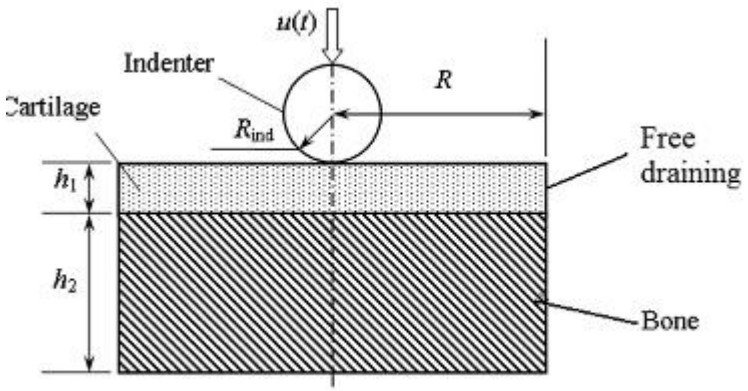


Fluid phase embedded in solid phase

Fluid flow through media is governed by
hydraulic permeability **k**
porosity **Φ**
Both depend on the deformation of the tissue

FE analysis (inverse ➔ simulate and find material properties by fitting deformation curve(exp) to simulation)

Example – FE simulations of a bi-phasic model



Qeng et al.2013

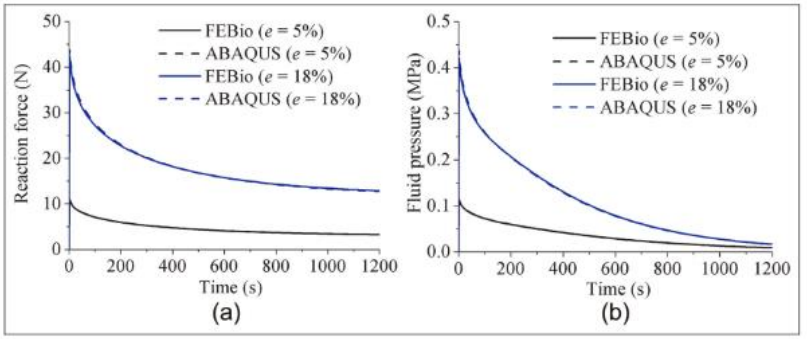


Figure 3. Comparison between FEBio and Abaqus of (a) the reaction force on the indenter and (b) the fluid pressure at the bottom of the cartilage (under the contact center) for the porous flat-ended model under different displacements.

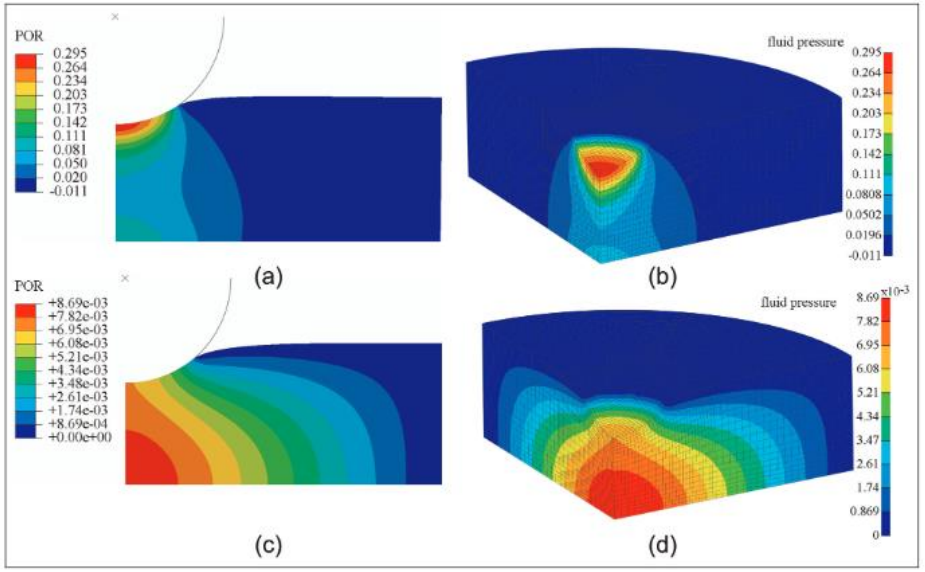
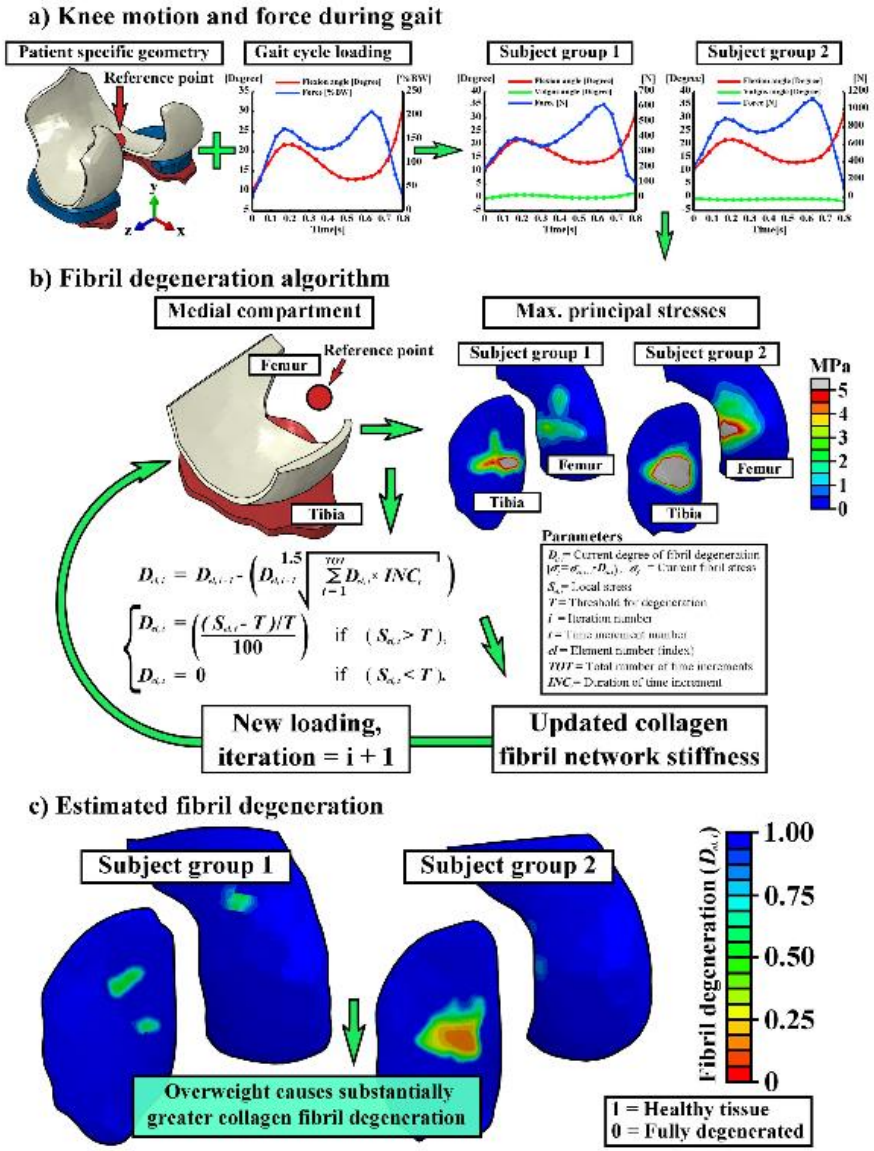


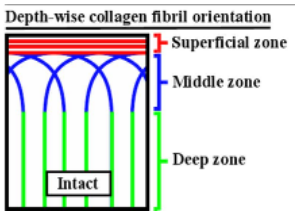
Figure 4. Fluid pressure distribution in the cartilage of the spherical-ended indentation model subjected to a load of 0.7 N, at 2 s, obtained by (a) Abaqus and (b) FEBio and for the same model at 1200 s, obtained by (c) Abaqus and (d) FEBio.

Include remodeling in material model (E(t))



Knee joint from imaging data +
Gait loading (gait biomechanics)

Fibril-reinforced cartilage material model with fibril stiffness depending on stress



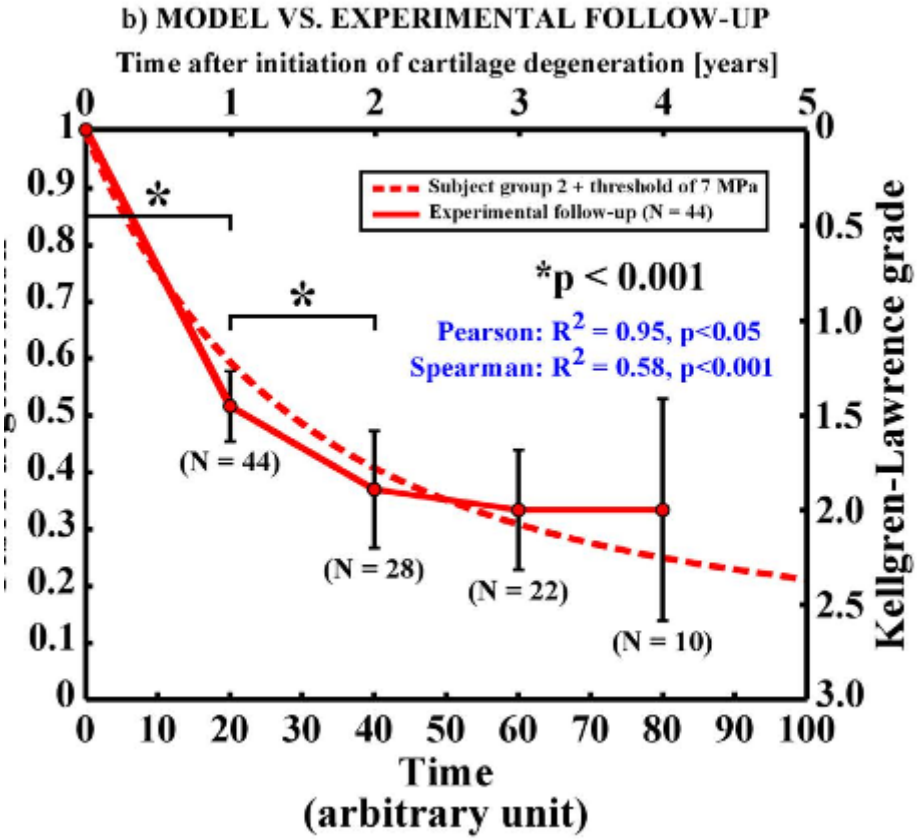
$$\boldsymbol{\sigma}_t = \boldsymbol{\sigma}_f + \boldsymbol{\sigma}_{nf} - p\mathbf{I}$$

Hyperelastic matrix (Neo-Hookean) +
pores +
viscoelastic fibrils(layer-dependent orientation)

Cartilage degeneration from FEM stress

Mononen et al. 2016

Model validation with clinical study



Comparison with clinical trial data of 44 patients with a 4-year follow up

Mononen et al. 2016

Summary material modeling

Material models to quantify material properties and further investigations in FEM simulations

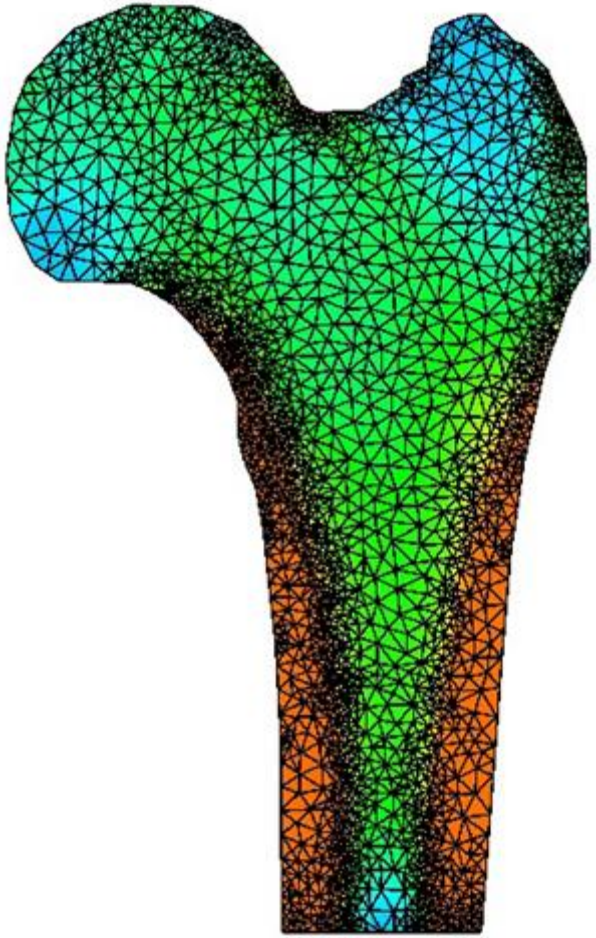
Choose what do you want to investigate (anisotropy, cyclic loading, creep, material failure...)➔ do experiments (uniaxial, biaxial, shear etc.) ➔ choose material model and fit parameters

Many factors play a role! It is difficult (not impossible) to include all of them in material model. E.g. hyperelasticity, anisotropy, remodeling, viscoelasticity, poroelasticity etc.

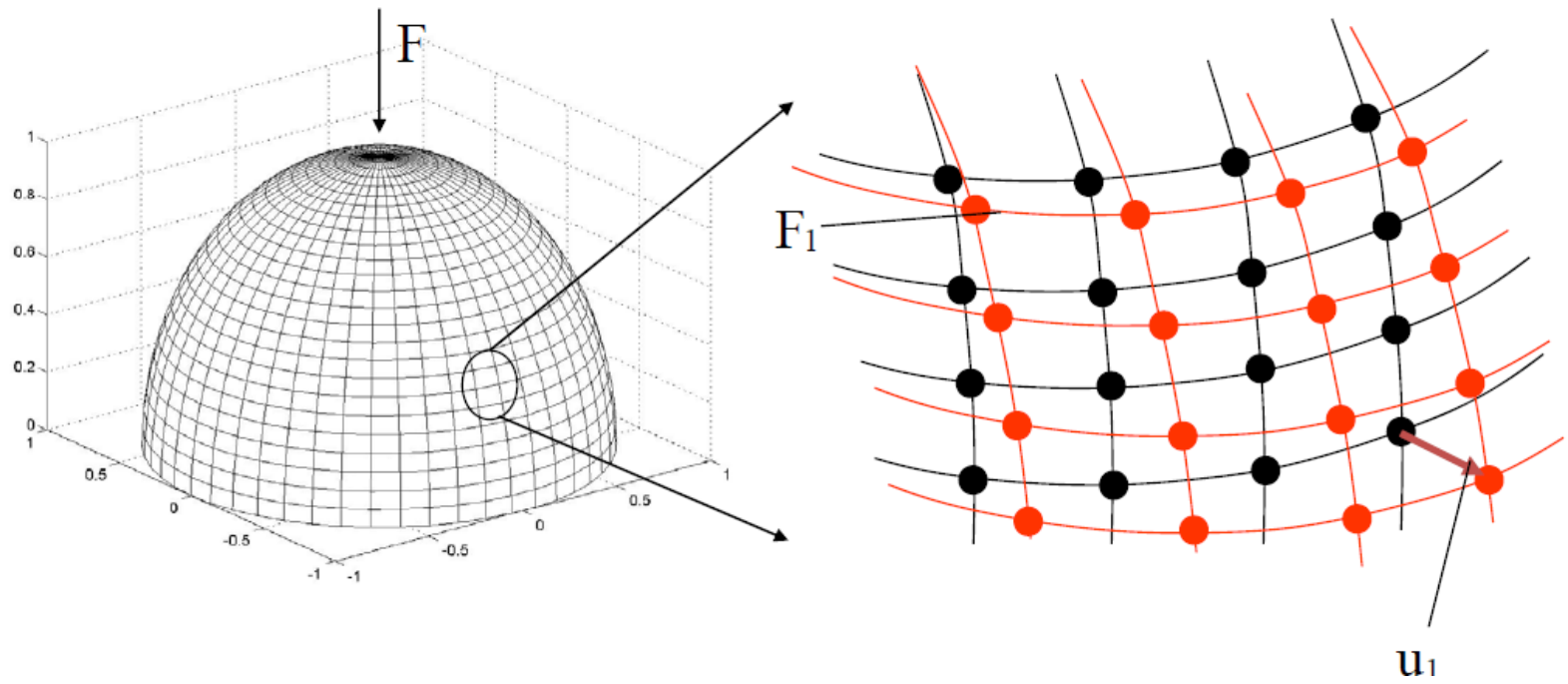
Implement model and validate with further experiments...

Questions?

FEM

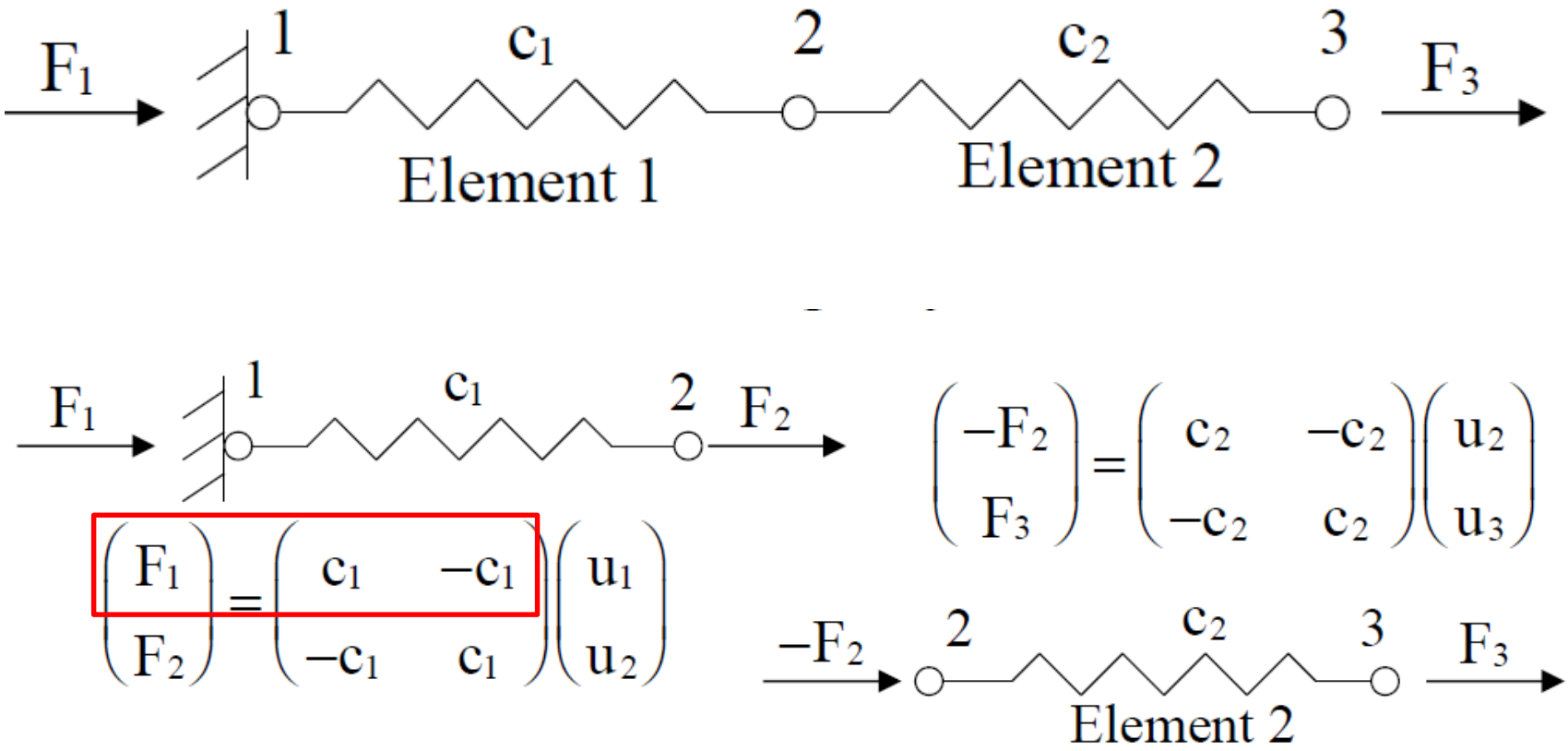


FEM Idea



Approach: Divide a complex geometry into a **mesh** of simple interconnected elements
Calculation of deformations in each cell with **approximate solutions**
Determination of total deformation after consideration of each cell deformation and existing interconnections

FEM simple example



Force at node 1 is spring constant 1 multiplied by the absolute length change of the spring.

$F_1=c_1*u_1-c_1*u_2$

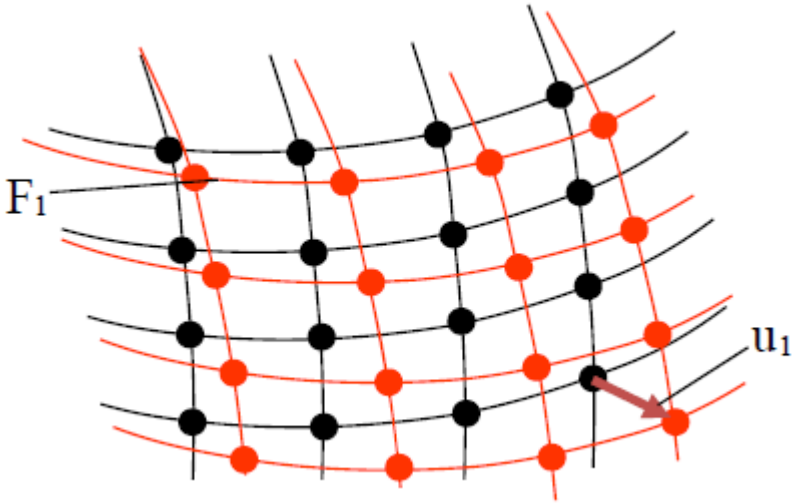
Stiffness matrix and boundary conditions

$$\begin{pmatrix} F_1 \\ F_2 \\ F_3 \end{pmatrix} = \underbrace{\begin{pmatrix} k_{11} & k_{12} & k_{13} \\ k_{21} & k_{22} & k_{23} \\ k_{31} & k_{32} & k_{33} \end{pmatrix}}_{\text{Stiffness matrix}} \begin{pmatrix} u_1 \\ u_2 \\ u_3 \end{pmatrix}$$

$$\begin{pmatrix} F_1 \\ 0 \\ F \end{pmatrix} = \begin{pmatrix} c_1 & -c_1 & 0 \\ -c_1 & c_1 & 0 \\ 0 & 0 & 0 \end{pmatrix} + \begin{pmatrix} 0 & 0 & 0 \\ 0 & c_2 & -c_2 \\ 0 & -c_2 & c_2 \end{pmatrix} \begin{pmatrix} 0 \\ u_2 \\ u_3 \end{pmatrix}$$

Boundary conditions (no movement of node 1, no spring 3) simplify the matrix

Application for real cases

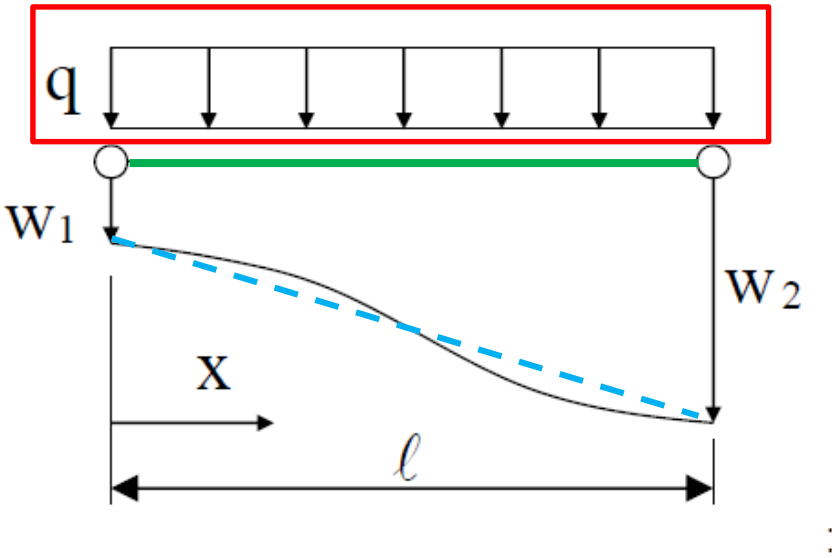


$$\begin{pmatrix} F_1 \\ \vdots \\ F_n \end{pmatrix} = \underbrace{\begin{pmatrix} k_1 & \cdots & \cdots \\ \vdots & \ddots & \vdots \\ \vdots & \cdots & k_n \end{pmatrix}} \begin{pmatrix} u_1 \\ \vdots \\ u_n \end{pmatrix}$$

Sparse (many entries are 0) matrix ➔ compute the inverse with numerical algorithms (PDEs)

Our elements are more complicated than simple mass-spring systems
➔ shear deformations, material models (viscoelasticity, big deformations etc...)
➔ take approximate solutions of each node deformation (next slide)

Approximation function for linear beam



On the beam **l** acts a distributed load **q**. The deformation is approximated with the linear function **w**.

After the approximation we have a system with a finite number of degrees of freedom (**DOF**s).
The system is solved (minimization of **internal energy** difference) numerically.

Summary FEM Theory

We **discretize** the geometry.

Express the **force/deformation** relationship with a system of **PDE**s.

This system has a number **DOF**s depending on the discretization approach.

System solution (matrix inversion) is done numerically.
We stop when the difference in our variables
(internal energy **U**, force **F**, deformation **u**) are below a **convergence** target.

More complicated cases need consideration of other factors (contact and shock, big deformations and remeshing, non-linear materials and numerical stability, fluid and solid interactions...)

The FEM method is well established (1950s-60s in the automotive and aerospace industry) and reliable commercial and open source solvers exist which allow user-friendly solutions of nowadays problems.

ANSYS ABAQUS COMSOL FEBiO Finite Elements for Biomechanics

FEM with focus on application

In the following the usual steps in a FE simulation are presented.

Focus is put on the reasoning and necessity of each step as well as possible pitfalls and problems that might come up during a simulation.

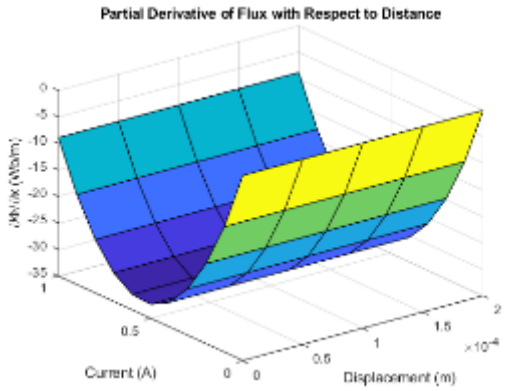
Commercial programs provide **huge** documentations with theory and application for every specific problem. **Tutorials** and workshops can be acquired from companies' websites.

The following should be considered as a **best practice** routine. What should be the mindset for a successful simulation.

Step 0: What's your problem?

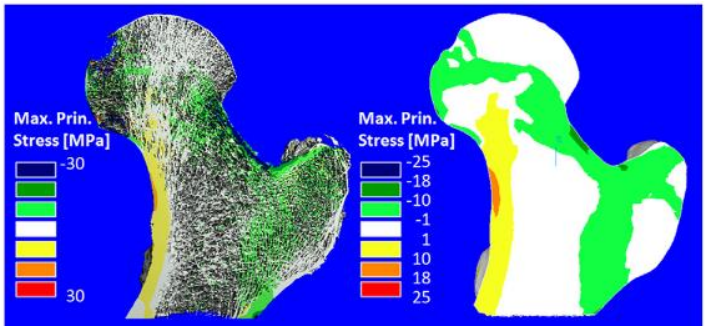
The research question defines the purpose and "style" of our simulation.

Parameterized design study



many simulations
high stability
low numerical costs

One specific case for mechanical insight



careful geometry extraction
realistic material modeling
high numerical costs

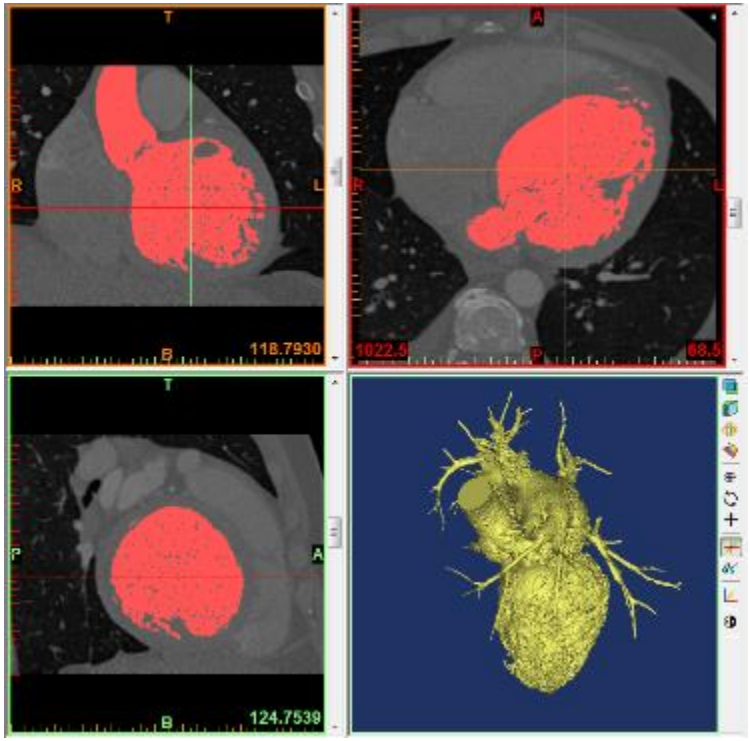
Marangalou 2012

Clinical study
Many uncertainties

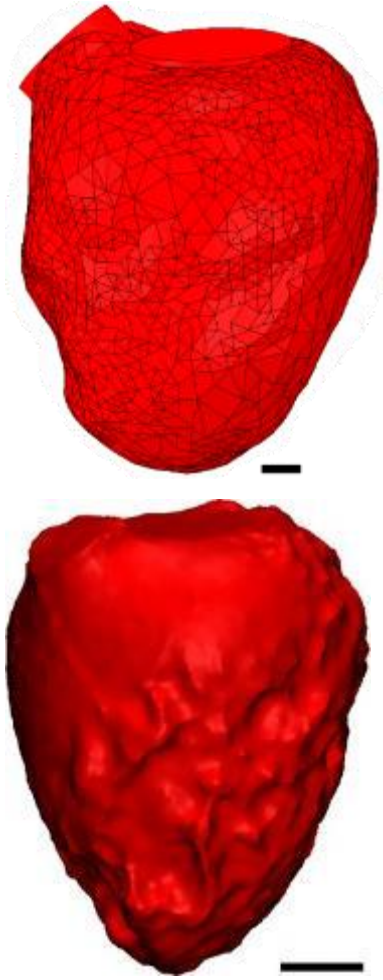
Method development
A lot of numerics

Step 1: Geometry creation/extraction

Especially after segmentation from imaging data the geometry has to be smoothed and/or trimmed accordingly



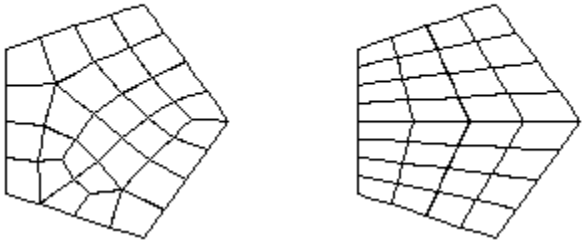
Liao et al. 2016



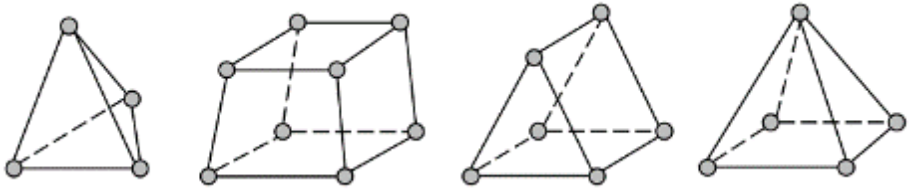
Step 2: Meshing

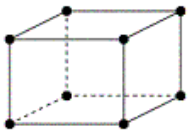
Decide on the mesh element type (free or mapped)

Figure 7-1 Free and mapped meshes

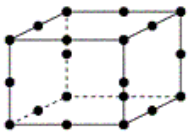


For complex geometries
often unstructured (free) tetrahedral meshes.
➔ Mesh size and numerical costs
➔ Element deformation (1 bad element can destroy entire simulation)
➔ Unstructured is much easier

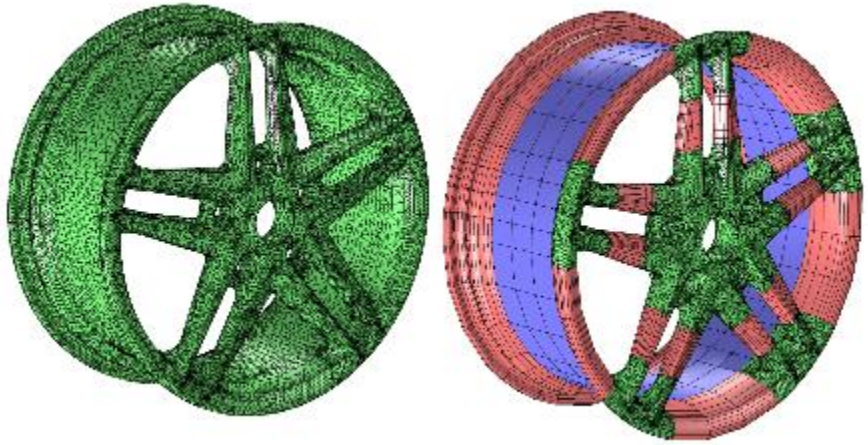




8-noded



20-noded



Step 3: Boundary conditions

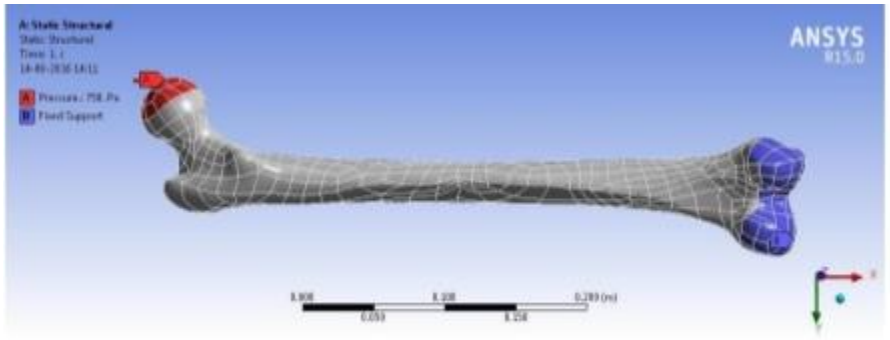


Fig. 6: Fixed Supports in the Lower Surface and Load at the Head Surface of Femur

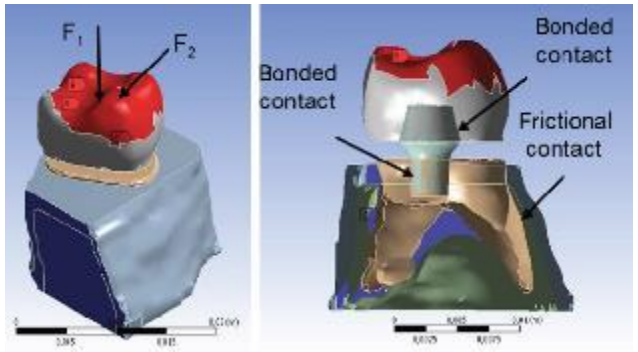
Force/pressure/momentum/displacement

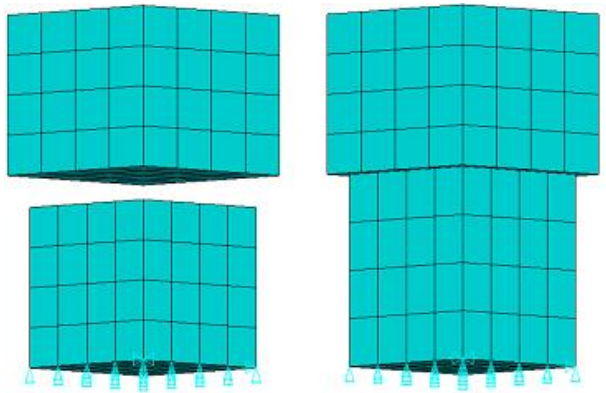
Locating/floating bearing

(Rotational) symmetry when possible.

Contact!

Contact pairs, force/deformation translation and penetration are often the reason for instabilities/slow simulations/problems.



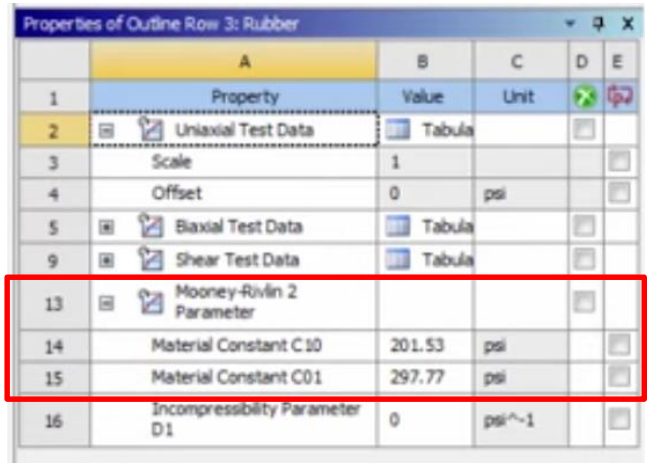


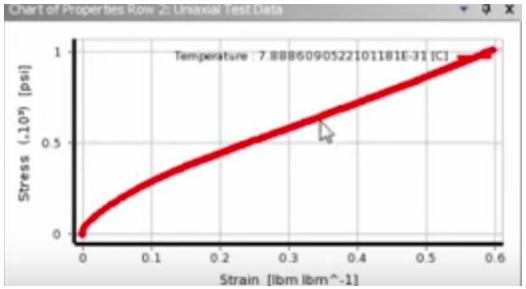
Gap closure for better stability

Step 4: Material models

Biological materials are very complicated. Material description and modeling is still ongoing. Based on the research question you should choose the phenomena you want to represent.

Especially **non-linearity** and **combinations of materials** with different properties (layers) can cause problems ➔ Better start simple, then switch to complicated material models if everything else works!





Material model choice and parameter fitting (like in lecture)

Step 5: Choice of solver

⊟	**Step Controls**	
	Number Of Steps	10.
	Current Step Number	2.
	Step End Time	2. s
	Auto Time Stepping	Program Controlled
⊟	**Solver Controls**	
	Solver Type	Program Controlled
	Weak Springs	Program Controlled
	Large Deflection	Off
	Inertia Relief	Off
⊞	**Restart Controls**	
⊞	**Nonlinear Controls**	
⊞	**Output Controls**	
⊞	**Analysis Data Management**	

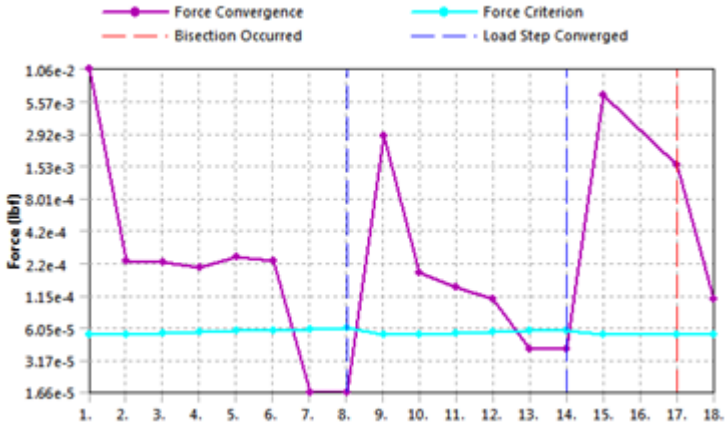
Stable solvers for various problems (**implicit/explicit**)

Time step size/duration of simulation

Convergence target

Under-relaxation factors (increased stability, increased calculation time)

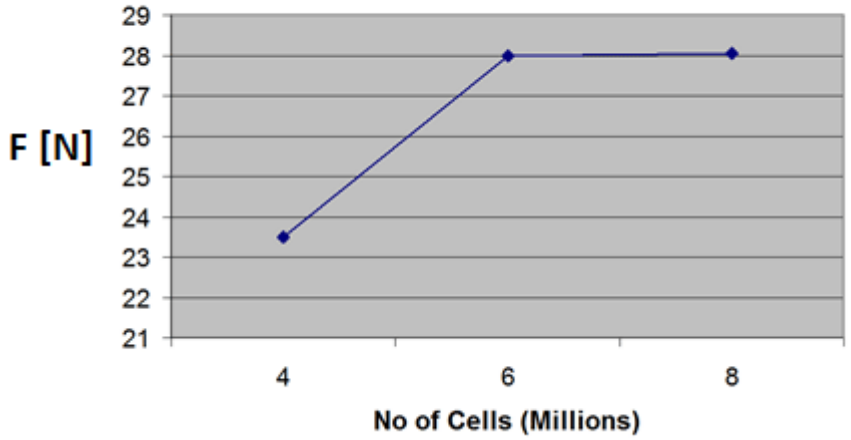
... a big world



Step 6: Independence studies

Our solution is an **approximation** of the initial geometry.

Guarantee that the solution does not depend on intrinsic factors like timestep, solver settings, b.c. and **mesh size**



Questions?

Imaging

Visual representation of human body parts or functions of organs and tissues

Analysis, Diagnosis, Intervention

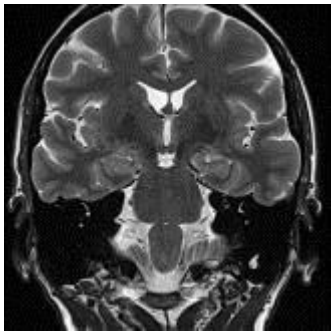
Clinician

Researcher

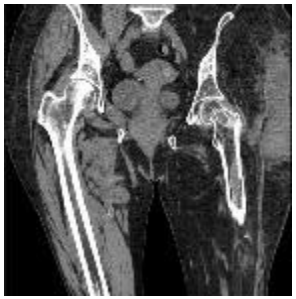
Engineer

Most of the methods are non-invasive

Imaging

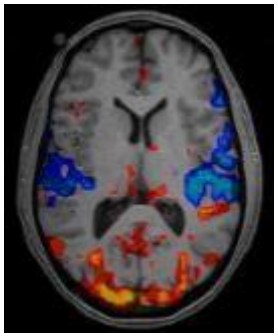


MRI

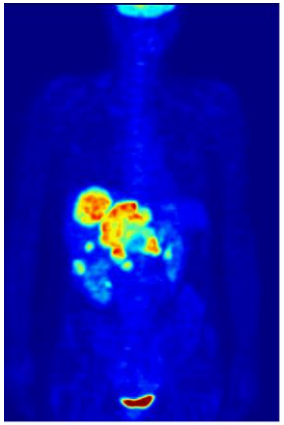


CT

Angiography

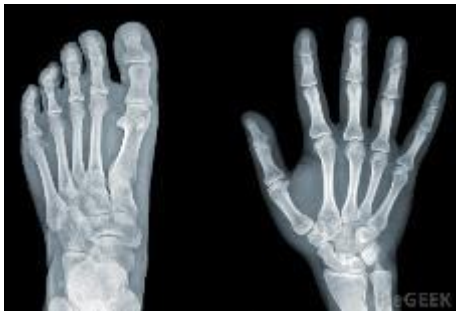


fMRI

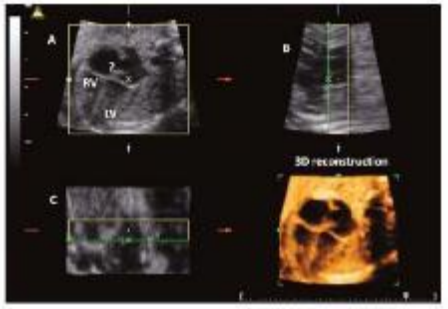


PET

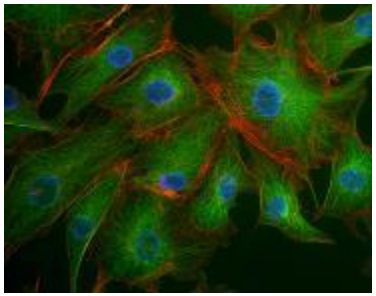
Medical Imaging Lecture
Prof. Αναγνωστάκης



X-ray

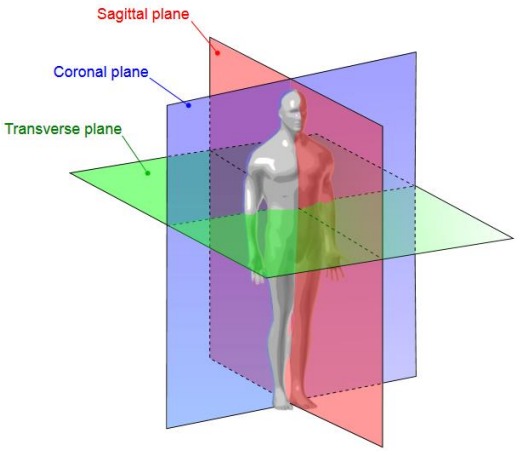


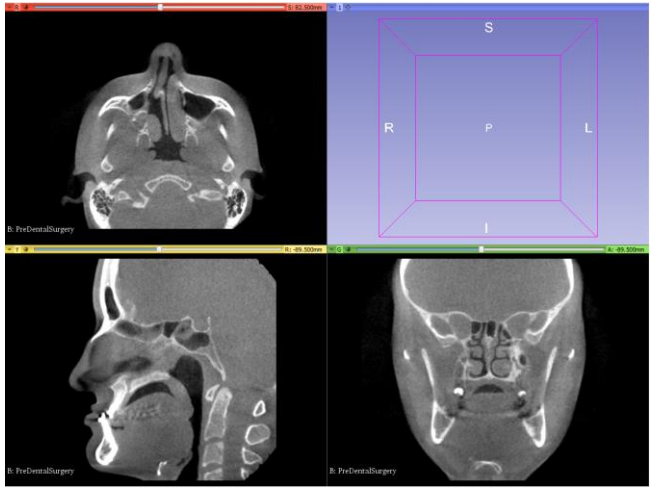
Echocardiography



Fluorescent imaging

CT (Computed tomography) working principle





Many X-ray scans taken from different angles to produce tomographic (sectional) images of 3D body

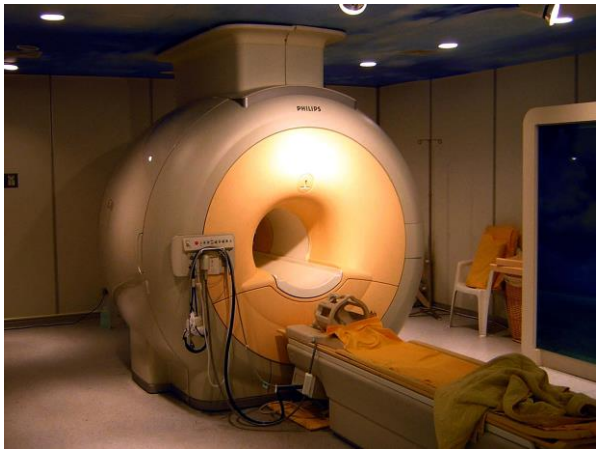
Different tissues absorb x-rays in a different way ➔ grayscale values

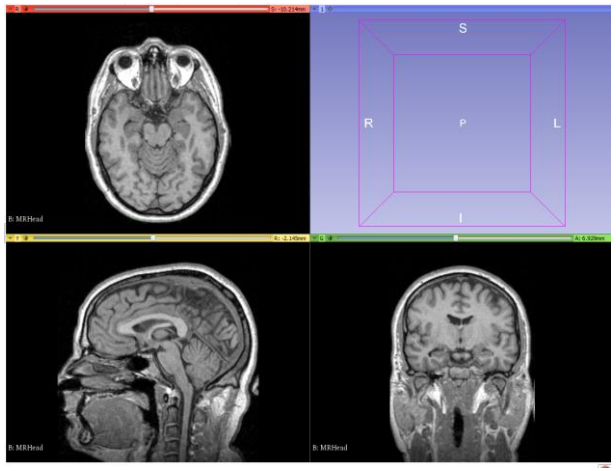
Many applications: hard tissues, tumors, cardiovascular..

Around 1-5mm slice thickness. μCT much smaller, just for research

Patient objected to radiation. **1-4x** yearly radiation exposure
➔ Benefit often outweighs

MRI (Magnet resonance imaging) working principle





Hydrogen atoms in tissues absorb and emit energy in an external magnetic field

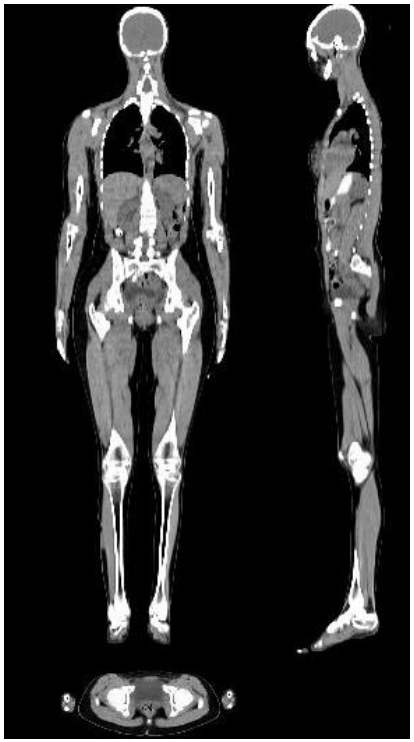
Different tissues react differently, water and fat locations➔ grayscale values

Many applications: soft tissues (muscles, ligaments, cartilage)
hard tissues, tumors, cardiovascular..

Around 0.2-5mm slice thickness. 4D MRI for time-dependent scans!

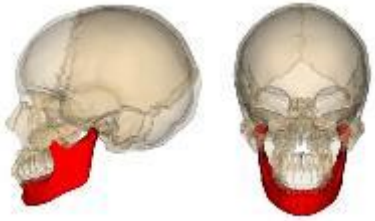
No radiation. Magnetic field ➔ not able to use for every patient. May be experienced as claustrophobic. Should not be overused ➔ Often as a 2. means of diagnosis

What do we need?

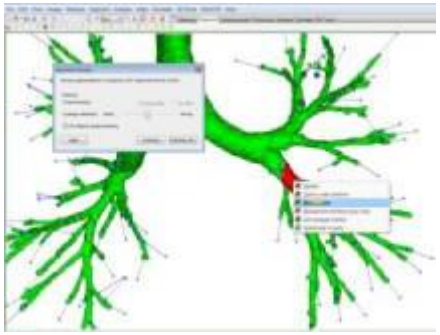


Further analysis (FEM,CFD etc.)
Models for in vitro studies
Device design

3D printing ➔ intervention planning



Patient-specific implants



Clinical studies

Gray-value images ➔ Triangular surface mesh (*.stl)

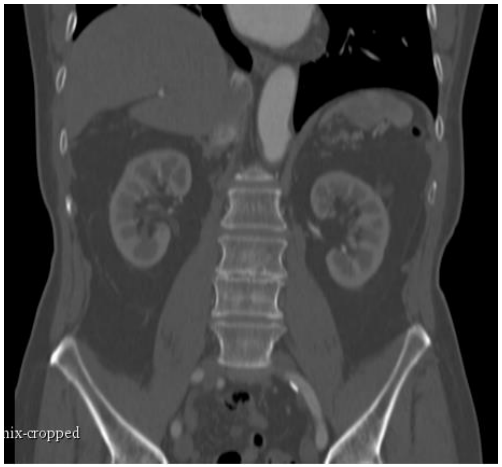
Segmentation – preamble

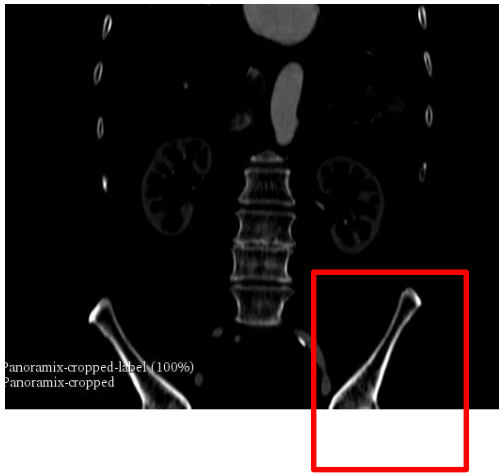
Segmentation is the partitioning of a digital image into multiple segments of interest.

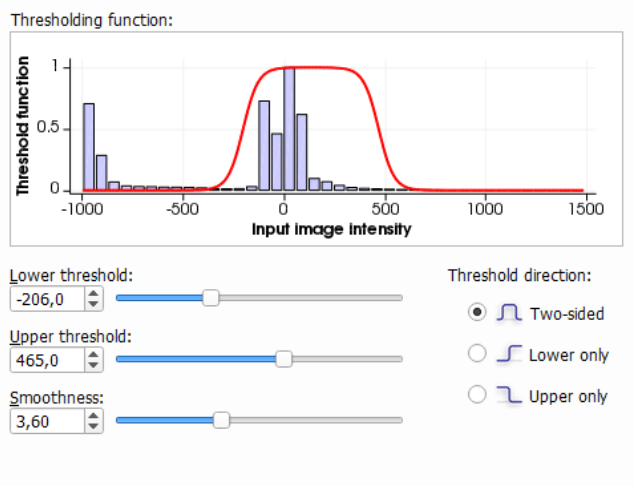
2D slices ➔ 3D surface

Theory of segmentation will not be covered in the following slides. In general there are many open-source and commercial programs which allow an efficient segmentation without knowledge of the background.

Image pre-processing - Thresholding



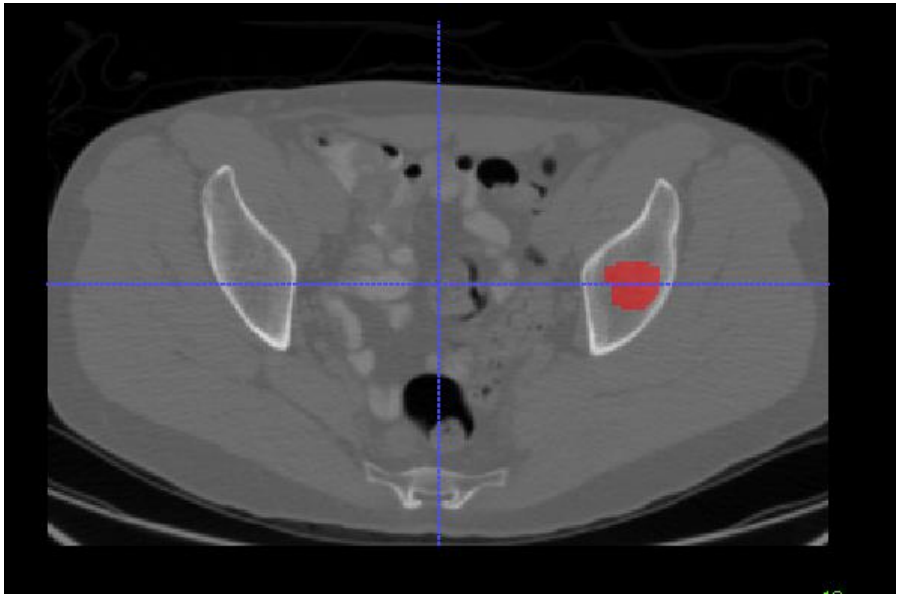


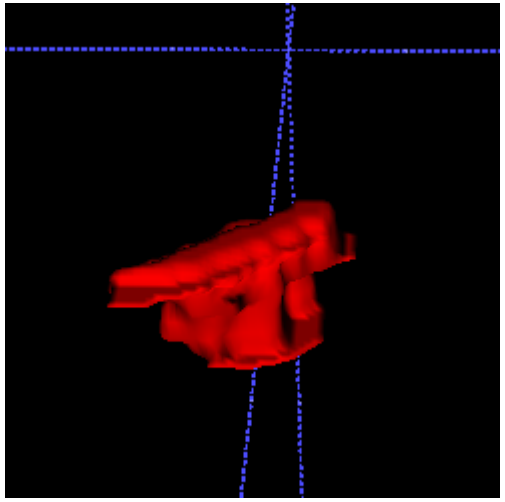


Number	Color	Name
0		background
1		tissue
2		bone
3		skin
4		connective tissue
5		blood
6		organ
7		mass
8		muscle
9		foreign object
10		waste

Different tissues produce different grayscale values. E.g. Bone vs. soft tissues. Increasing the intensity/contrast of the respective intensity values helps to distinguish the tissues. Some tissue thresholds are already stored inside the software.

Manual segmentation

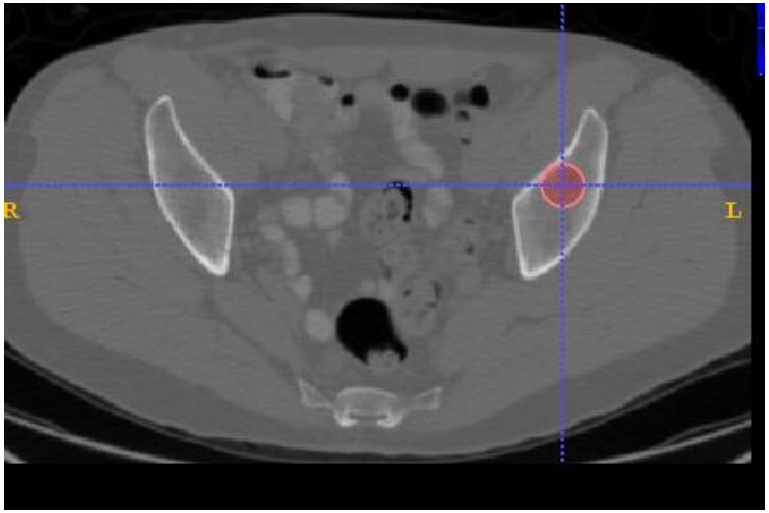


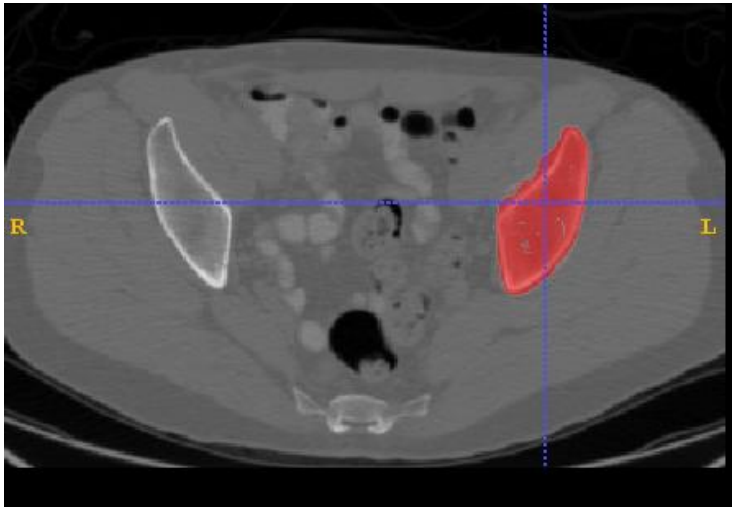


Very straightforward procedure. Marking the regions of interest (red paint) on each slice and creating 3D model them.

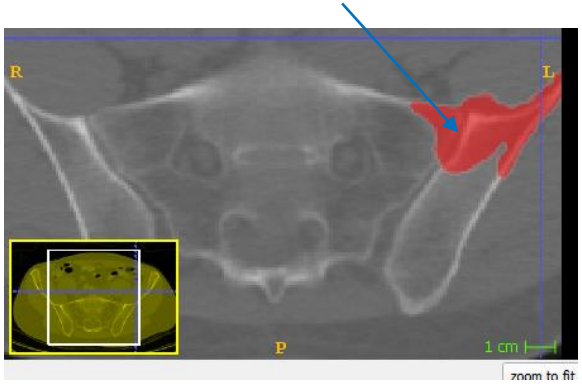
Semi-automatic segmentation

Basic idea: Place seeds/sources at the parts of interest. Then let the regions grow until the boundary of the part of interest is reached.





However mistakes like this can appear



Pre-processing and modification of segmentation parameters based on application case is important. Also the **knowledge** of the geometry!!!

Semi-automatic segmentation

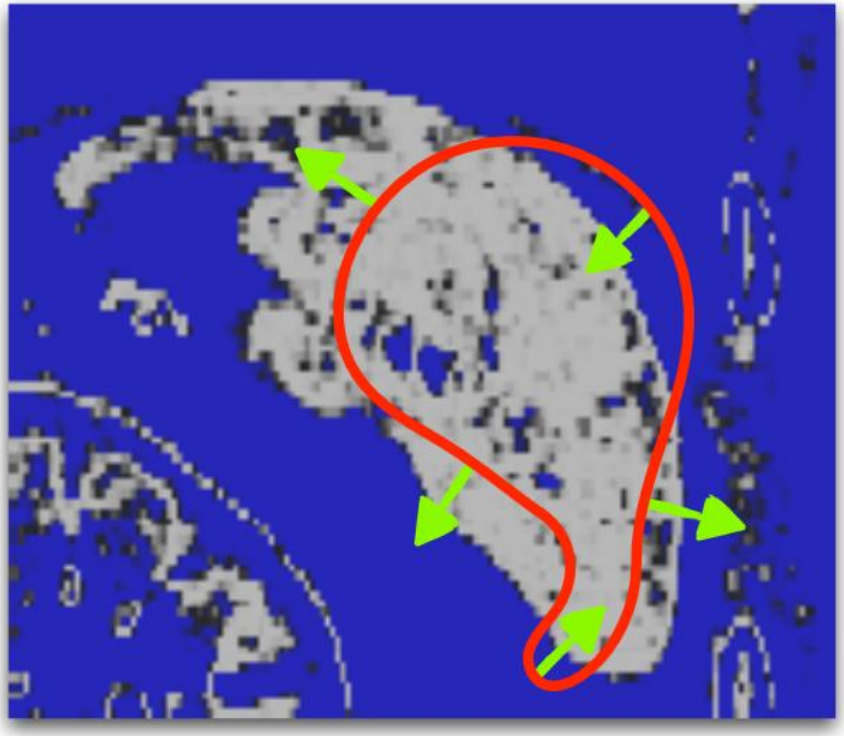
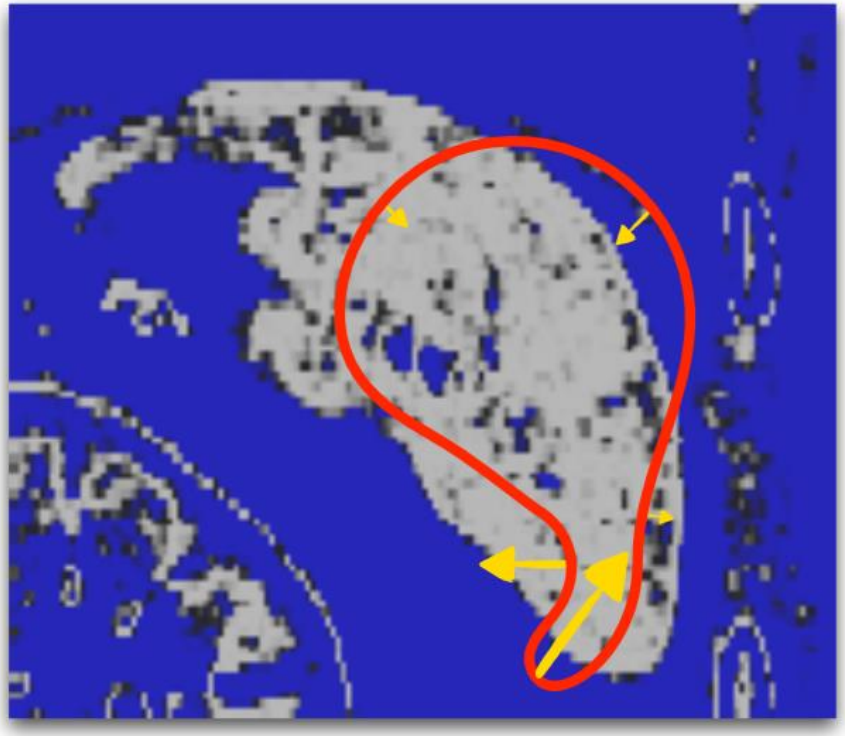
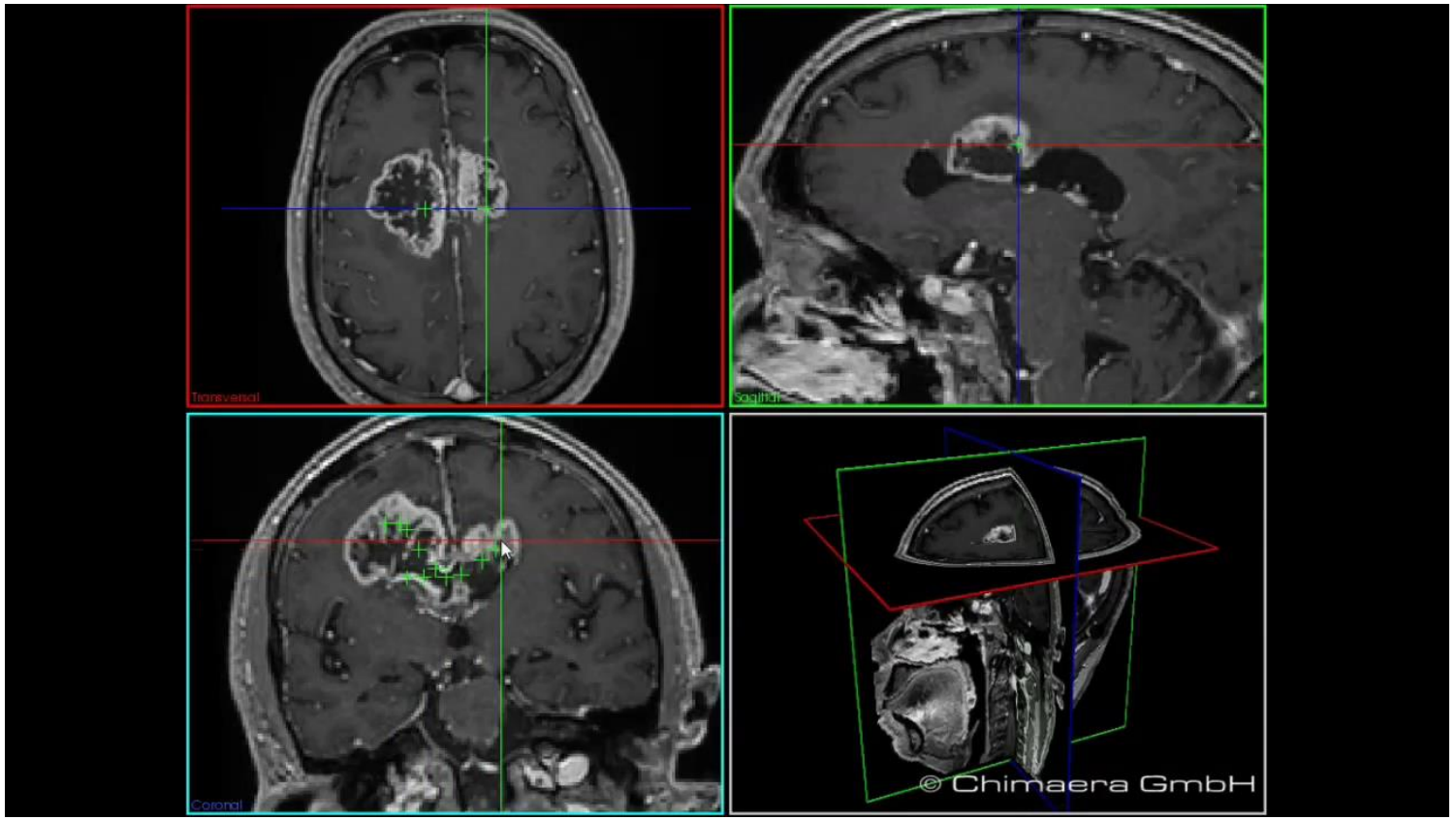


Image Force:
outward over foreground, inward over background



Smoothing Force:
strongest in places of high curvature

Semi-automatic segmentation



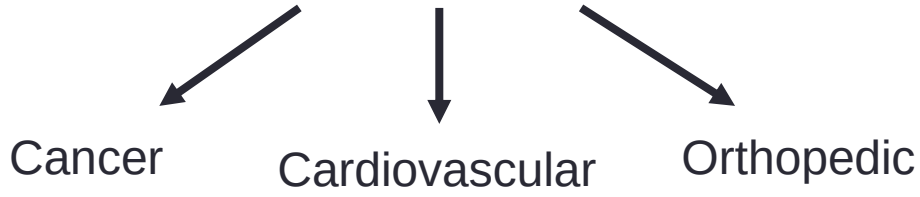
Careful pre-processing is important!
After model creation ➔ post-processing (smoothing, cut-planes etc.)

Automatic segmentation

Very active area of research and development

Artificial intelligence and **machine learning** are supposed to overcome the limitations of current algorithms and problems like **inter-**observatory and **intra-observatory** variations

Huge medical impact and databases existing (hospitals)



Questions?

Summary

Investigations in biomechanics include various fields such as:

Signal processing, image analysis, general mechanics, multi-body dynamics, heat and mass transfer, fluid dynamics, statistics etc.

There are **no biological constants**, every parameter experiences huge variations (stiffness, size, weight...) and changes over time (remodeling).

The engineering approach of understanding and modifying complex systems reaches its limits ➔ however might be still better than the clinical ‘trial-and-error’ approach.

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